

THE EFFECT OF GRADE ON THE BIOMECHANICS OF DOWNHILL
RUNNING IN FEMALE DISTANCE RUNNERS

A THESIS

SUBMITTED TO THE GRADUATE SCHOOL
IN PARTIAL FULFILLMENT OF THE REQUIREMENTS

FOR THE DEGREE

MASTER OF SCIENCE

BY

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BALL STATE UNIVERSITY

MUNCIE, INDIANA

MAY 2017

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Declaration

The work presented in this thesis is, to the best of my knowledge and belief, original, except as acknowledged in the text, and the material has not been submitted, either in whole or in part, for a degree at this or any other university.

Meredith D. Wells

Date

Acknowledgements

I would like to thank Dr. Henry Wang for helping me throughout the thesis process. Dr. Wang helped me from the start of the writing process through the final submission of my thesis document and beyond. His guidance was exceedingly helpful, and I appreciate the time and effort he put in to helping me. The experience and knowledge that I have gained from his work has been invaluable to me. I would also like to thank my other committee members, Dr. Clark Dickin and Dr. Jennifer Popp. I appreciate your thoughtful comments and critiques that facilitated my finishing the thesis project successfully and to the best of my ability. Thank you for investing your time and commitment to make my research better.

Abstract

THESIS: The effect of grade on the biomechanics of downhill running in female distance runners

STUDENT: Meredith Wells

DEGREE: Master of Science

COLLEGE: College of Health

DATE: May, 2017

PAGES: 103

Hill running, both up and down, is often used as a foundational training mechanism to build strength and speed. Distance runners in particular are at an increased likelihood of encountering steep hills during long training runs. There is limited research available regarding downhill running, and there is no research available on the biomechanics of females during downhill running, or on the biomechanics of steep hill running. The purpose of this study was to quantify the biomechanics of downhill running at four different grades compared to level in female distance runners, and to determine the potential injury risk when running downhill. Fifteen healthy, female distance runners between the ages of 18 and 35 who ran a minimum of 15 miles per week with no lower extremity injuries participated in this study. Participants ran on a force-instrumented treadmill at 4.0 m/s for 1-3 minutes at 0%, -5%, -10%, -15%, and -20% grades, with 3-5 minutes of rest between each condition. Study findings indicated increased ground reaction forces and loading rates, greater power absorption at all three lower extremity joints, increased range of motion at the knee, decreased hip and knee flexion and increased trunk extension at initial contact, and increased knee and hip adduction moments. The results from this study indicate that there is a potentially greater risk for overuse injury when running downhill

compared to running on a level surface. Individuals should be aware of these findings when planning and implementing training programs so that overuse injuries may be avoided.

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Chapter 1

The Problem

Introduction

Running is a popular activity among individuals of all ages, ethnicities, and economic backgrounds. Vast numbers of individuals will run for recreational enjoyment, health and fitness benefits, and/or to fuel their competitive nature. For years, it has been known that distance running yields numerous health benefits, including cardiovascular and cardiorespiratory fitness, and muscular strength and endurance. Physical fitness activities, such as running, can also aid in the prevention of stroke, heart disease, diabetes, and osteoporosis (Nicholl, Coleman, & Brazier, 1994). Running continues to grow in popularity as the medical focus moves from treating disease to preventing disease (Nicholl et al., 1994; van Gent et al., 2007). Long distance running in particular is practiced by many people because it requires little to no skill, is low cost, and can be done anywhere. For individuals who wish to compete, races are available to all ages, on all terrains, at varying distances. The number of miles runners train per week will vary greatly depending on the individual goals and objectives of the participants. Long distance runners have an increased likelihood of running up and down hills while out on long runs, and competitive runners use hill running, both up and down, as a key training technique to build strength and speed (Paradisis, Bissas, & Cooke, 2009). Additionally, many distance runners train on trails,

which often introduces more inclines and declines than running on paved surfaces. The hills encountered by trail runners can be significantly greater than those encountered when running on roads.

Despite the numerous benefits of running, there is also a significant chance for injury at some point in a runner's career (Marti, Vader, Minder, & Abelin, 1988; van Gent et al., 2007). Anywhere from 27 to 79 percent of distance runners will experience an overuse injury over the course of a year (Lun, Meeuwisse, Stergiou, & Stefanyshyn, 2004; Marti et al., 1988; Maughan & Miller, 1983). Exercise related leg pain is a common occurrence in distance runners. In a survey of collegiate cross country teams, 60 percent indicated they had experienced lower extremity pain at some point in their running career, and 50 percent of these reported that the pain caused them to miss practice and/or it negatively affected performance (Reinking, Austin, & Hayes, 2007).

Musculoskeletal overuse injuries occur due to forces being applied to a structure over a period of time that is beyond the capabilities of that structure, such as muscle, tendon, or bone. Repetitive forces that are small in magnitude, but high in number are likely to cause to an overuse injury. There are multiple factors that can cause overuse injuries including training errors, lower extremity malalignment, and biomechanics (Hreljac & Ferber, 2006). However, not all of these factors can be avoided.

Kinetic variables such as high vertical ground reaction forces (GRFs), and high impact loading rates have been linked to overuse injuries in runners (Hreljac, Marshall, & Hume, 2000). Tibiofemoral compressive forces can be as great as 10 times body weight, patellofemoral compressive forces can be as great as 7 times body weight, compressive forces at the ankle and lower leg can be over 11 times body weight, and GRFs can be as great as 3 times body weight

(William Edwards, 2009; Flynn & Soutas-Little, 1995; Messier et al., 2008; Scott & Winter, 1990). Forces of this magnitude, without proper recovery time, are likely to result in an overuse injury. In addition, decreased ankle pronation, varus knee alignment, and high Q angles have been seen in individuals with patellofemoral pain syndrome (PFPS), iliotibial band friction syndrome (ITBFS), and stress fractures (Hreljac et al., 2000; Taunton et al., 2002). Female runners also tend to be at an added risk of injury when compared to males primarily due to a greater Q angle, which in turn leads to altered biomechanics (Gwinn, Wilckens, McDevitt, Ross, & Kao, 2000; Horton & Hall, 1989; Knapik et al., 2001; Myklebust, Maehlum, Holm, & Bahr, 1998; Reinking et al., 2007; Taunton et al., 2002; van Gent et al., 2007).

Some of the most common overuse running injuries include PFPS, ITBFS, and stress fractures (Taunton et al., 2002). The most likely causes of PFPS are high GRFs, high knee joint moments, a greater Q angle, and excessive frontal plane motion (Ferber, Davis, & Williams III, 2003; Horton & Hall, 1989; Mizuno et al., 2001; Powers, 2003). PFPS is often heightened by downhill running because it causes damage to the flexor and extensor role of the knee (Rolf, 1995). The most likely causes of ITBFS are excessive frontal and transverse plane movements (Hamill, Miller, Noehren, & Davis, 2008). Downhill running has been reported to increase irritation in individuals with ITBFS because it involves landing in a more extended knee position (Hamill et al., 2008). Distance runners are highly prone to stress fractures due to the repetitive loading of the lower extremity (Edwards, Taylor, Rudolphi, Gillette, & Derrick, 2009). GRFs are thought to be the primary cause of stress fractures (Lorimer & Hume, 2014). All of these injuries can lead runners to being forced to take time off in order to let the injured structures heal. This recovery time can range from a couple of days to a couple of months depending on the severity of the injury. An injury, depending on the severity, may also come with a financial burden.

Studies have indicated that downhill running may be more likely to cause an injury than level or uphill running (Buczek & Cavanagh, 1990; Burr et al., 1996; Gottschall & Kram, 2005; Milner, Ferber, Pollard, Hamill, & Davis, 2006; Paradisis & Cooke, 2001; Telhan et al., 2010; Yokozawa, Fujii, & Ae, 2005). Running downhill is not unlike repeated single leg landings. Single leg landings require eccentric contraction of the muscles in order to slow down the rate of energy absorption. The forces produced during this lengthening can be extremely high resulting in muscular fatigue and the breakdown of collagen, which can lead to an increased risk of injury (Brown, Day, & Donnelly, 1999; LaStayo et al., 2003). Additionally, eccentric muscle activity involves a high amount of force being applied to fewer motor units, which results in increased delayed onset muscle soreness (DOMS) (Cai et al., 2010).

The height and distance from which the landing occurs can also result in altered mechanics. A greater vertical height and/or a greater horizontal distance from which the landing occurs results in a greater vertical GRF, decreased knee flexion, and decreased hip flexion (Ali, Rouhi, & Robertson, 2013). Females tend to land in a more dorsiflexed position which contributes to a greater GRF as well (Ali et al., 2013). When running downhill, previous research has shown that runners tend to land heel first, or in a dorsiflexed position, with decreased hip and knee flexion, similar to what has been found to occur with single-leg landings (Buczek & Cavanagh, 1990; Gottschall & Kram, 2005; Yokozawa et al., 2005). The primary difference between single leg landings and downhill running is the running surface. Single leg landings generally occur on even surfaces, while when running downhill the landing occurs on an uneven surface. This may introduce a greater risk for injury. Most running injuries likely occur downhill, however, the mechanics of downhill running are not well understood, especially when it comes to steeper running grades.

Although numerous studies have been conducted on level running, there is limited research available regarding downhill running. Previous research has indicated that downhill running produces different biomechanics than running on level and uphill surfaces (Buczek & Cavanagh, 1990; Cai et al., 2010; Gottschall & Kram, 2005; Mizrahi, Verbitsky, & Isakov, 2001; Yokozawa et al., 2005). However, the studies that have been done on downhill running have been concentrated at low declines. It is unknown what happens biomechanically when the decline becomes greater. We do not know whether the risks associated with downhill running plateau at a certain grade, or whether the risks become more, or potentially less, severe with increasing declines. It is important to understand the biomechanics associated with steep downhill running grades because this running condition occurs frequently during distance running.

Purpose

The primary purpose of this study was to quantify the differences in mechanics and muscle activity for downhill running at the four different grades compared to level running in female distance runners, and to determine if there was a dose-response relationship between the grade and the biomechanical variables. The secondary purpose of this study was to analyze female distance runners' lower extremity joint kinematics, kinetics, spatiotemporal parameters, and muscle activity in order to determine the potential risk for injury.

Significance

The results from this study may be used to help enhance the current biomechanics literature on running, specifically with regards to downhill running. Downhill running is common when it comes to distance running training and racing, and data is lacking when it comes to declines greater than -10%. The findings from this study may also be used to determine

if steep declines result in running patterns that are susceptible to an increased risk of injury, or if the risk for injury either plateaus or decreases with increasing decline. Additionally, this information can help individual distance runners, and coaches to develop training programs that may help avoid sustaining a running related injury.

Hypotheses

There were three general research questions I wanted to answer. 1) How will the vertical ground reaction forces, along with the braking and propulsive ground reaction forces change when running downhill at different declines? 2) Will the muscle activity in the lower extremity change as the grade becomes steeper? 3) Will there be differences in the spatiotemporal parameters, kinematics, and kinetics among the different grades? My hypotheses were 1) increased vertical ground reaction forces, and increased braking ground reaction forces and decreased propulsive forces. 2) Changes in muscle activity with changing grade. 3) Decreased stride rate and increased step length, greater hip and knee joint extension at ground contact, and greater joint moments at the ankle and knee.

Limitations

Limitations for this study were footwear preference, and running surface. All participants were given identical athletic shoes to wear in order to eliminate the possibility of altering the results due to differences in shoe condition or inserts. Being unfamiliar with the type of shoes worn could have introduced some differences. However, because everyone was in the same kind of shoe, this was hopefully avoided. The study was also conducted in a controlled laboratory setting on a force-instrumented treadmill. Treadmills have been found to be valid instruments for running data collection, so this should not have affected the ability to relate the results to real-world settings.

Delimitations

Only healthy individuals between the ages of 18 and 45 that are free of chronic musculoskeletal and/or neurological conditions were included in this study. This study was also limited to only female runners who train a minimum of fifteen miles per week. The participants were in standardized footwear, which could have potentially altered the normal stride patterns of the individuals. The running speed was set at a constant speed of 4.0 meters per second for each grade, which may have varied the degree of difficulty felt at each level, as well as the difficulty felt by each runner individually. The study was also conducted in a controlled laboratory on a force-instrumented treadmill.

Summary

Distance running is a popular activity that individuals from all backgrounds partake in. However, limited information is available regarding downhill, especially when it comes to running downhill at steep declines, which can occur during long training runs and races. This has limited the ability of coaches and runners to fully understand the cause of running overuse injuries. The primary purpose of this study was to quantify the differences in mechanics and muscle activity for downhill running at the four different grades compared to level running in female distance runners, and to determine if there was a dose-response relationship between the grade and the biomechanical variables. The secondary purpose of this study was to analyze female distance runners' lower extremity joint kinematics, kinetics, spatiotemporal parameters, and muscle activity in order to determine the potential risk for injury.

Chapter 2

Review of the Literature

Introduction

Running is a popular sport enjoyed by individuals of all ages for recreation, health and fitness, and/or competition. Long distance running in particular is practiced by many people as a result of its convenience. With the growing interest in disease prevention running continues to grow in popularity due to the health benefits it elicits (van Gent et al., 2007). For individuals with the competitive spirit, races are available to all age groups, on all terrains, at distances ranging from one kilometer road races to 100 mile ultra-marathons. The number of miles distance runners will train each week differs significantly, depending on the individual's goals and objectives. The running surface can vary as well, from synthetic tracks, to asphalt roads, to hilly dirt or gravel trails, to mountainous terrain. Long distance runners have an increased likelihood to be running up and down hills while out on long runs, and many competitive distance runners use hill running, both up and down, as a foundational training mechanism.

Although the benefits of running are numerous, there is also significant chance for injury at some point in a runner's career (Marti et al., 1988; van Gent et al., 2007). This literature review will focus on the kinematics, kinetics, and muscle activity of running on different grades, and how the differences may or may not relate to injury in long distance runners. More

specifically, this literature review will provide insight into three common running injuries along with some of the most common mechanisms of injury, in addition to the joint angles and moments, ground reaction forces, and muscle activity seen in horizontal, uphill, and downhill running.

Validity of instrumented treadmills for running data collection

The majority of biomechanical running data must be collected in laboratory settings in order to have access to the appropriate instruments, such as infrared cameras to collect motion data and force plates to collect ground reaction forces and moments. Therefore, it is important to be able to extend research performed in laboratory settings to real world, out of doors, settings. Riley, Dicharry, Franz, Croce, Wilder, and Kerrigan (2008) sought to do just this. The purpose of their study was to compare the kinematics and kinetics of running on an instrumented treadmill to over ground running. They found that the stride time and stride length was shorter when running on a treadmill, and that both the peak propulsive anterior-posterior and peak medial-lateral ground reaction forces were lower in treadmill running. However, it was determined that the kinematics and kinetics of running were very similar between the two conditions. Thus, they concluded that over ground running and instrumented treadmill running are comparable. This finding validates the use of laboratory conducted research to study running biomechanics.

Common risk factors associated with overuse injury in runners

Running is one of the most common activities in which musculoskeletal overuse injury of the lower extremity occurs (Hreljac et al., 2000). Various studies have found that over the course of a year anywhere from 27 to 79 percent of distance runners will experience an overuse injury (Lun et al., 2004; Marti et al., 1988; Maughan & Miller, 1983), and the majority of overuse injuries occur at or below the knee (Maughan & Miller, 1983; Rolf, 1995; Taunton et al., 2002;

van Gent et al., 2007). Exercise related leg pain (ERLP) is a common occurrence in collegiate cross country runners (Reinking et al., 2007). A series of questionnaires completed by five collegiate cross country teams, spanning all collegiate athletic divisions, indicated that approximately 60 percent of these runners experienced ERLP at some point in their running career. Of these, 50 percent reported it had caused them to miss practice at some point, and 60 percent reported that it had negatively affected performance (Reinking et al., 2007).

Musculoskeletal overuse injuries arise as a result of combined fatigue over an extended period of time beyond the capabilities of a specific structure such as muscle, tendon, or bone (Hreljac & Ferber, 2006). A large number of repetitive forces that are small in magnitude applied to the structure without sufficient recovery time will likely result in this kind of injury. There are multiple factors that are capable of causing overuse injuries in runners including training errors, anthropometrics, lower extremity malalignment, and biomechanical influences (Hreljac & Ferber, 2006).

Although there are several risk factors that could potentially lead to an overuse injury in runners, not all of them can be avoided. For example, anthropometric and lower extremity malalignment risk factors such as Q angle (Mizuno et al., 2001) and leg length discrepancy (Hreljac & Ferber, 2006), cannot be modified. Whereas training distance and intensity and select biomechanical factors can be altered. This section will focus on the biomechanical risk factors associated with running injuries. Biomechanical variables can be broken down into kinematic and kinetic variables. Kinematic variables describe the motion of a point or body. Kinetic variables describe the causes of such motion. Kinetic variables such as high vertical ground reaction forces (GRFs) and high vertical impact loading rates have been linked to many overuse injuries in runners (Hreljac et al., 2000). Tibiofemoral compressive forces have been found to be

as high as 10 times body weight (Messier et al., 2008), peak patellofemoral forces to be between 4 and 7 times body weight (William Edwards, 2009; Flynn & Soutas-Little, 1995; Messier et al., 2008), and peak GRFs to be between 2 and 3 times body weight (Scott & Winter, 1990).

Additionally, it has been demonstrated that the force applied to the Achilles tendon during stance is greater than 6 times body weight, and the compressive forces at the ankle and lower leg are over 11 times body weight (Scott & Winter, 1990). The accumulation of forces of this magnitude without sufficient recovery time has a high likelihood of resulting in an overuse injury.

Kinematic variables such as decreased ankle pronation have also been linked to running overuse injuries (Hreljac et al., 2000). Pronation allows for the impact forces to be attenuated over a longer period of time, which serves as a protective mechanism during running (Hreljac & Ferber, 2006; Subotnick, 1985). Further, varus knee alignment has been seen in individuals with patellofemoral pain syndrome, iliotibial band friction syndrome (ITBFS), and tibial stress fractures, and genu valgum and high Q angles have been seen in individuals with PFPS and patellar tendinopathy (Taunton et al., 2002).

Additionally, the female sex tends to be at a greater risk of sustaining running related injuries compared to males. In a synthesis of 17 articles relating to lower extremity injuries in long distance runners, the only significant association over all lower extremity injuries was with female sex (van Gent et al., 2007). An analysis of 2002 running injuries found that 8 percent more females were injured than males (Taunton et al., 2002). This same study illustrated that females were more likely to sustain injuries such as PFPS and ITBFS, while men were more likely to experience plantar fasciitis, meniscal injuries, Achilles tendinopathy, and osteoarthritis of the knee. In addition, females are five times more likely to sustain an anterior cruciate ligament (ACL) injury than males in handball (Myklebust et al., 1998), 9 times more likely to

become injured during varsity athletics, 7 times more likely to become injured in intramural athletics, twice as likely to become injured during military training (Gwinn et al., 2000), and twice as likely to become injured during basic combat training (Knapik et al., 2001).

Additionally, females have been found to sustain stress fractures at a rate almost three times that of males (Reinking et al., 2007).

Sex differences that may lead to more injuries in female runners.

As stated earlier, men and women tend to accrue different injuries (Taunton et al., 2002), which is likely due to anatomical and biomechanical differences (Ferber et al., 2003; Taunton et al., 2002). Anatomical differences between males and females predispose females to certain injuries, including PFPS, ITBFS, and tibial stress fractures (Taunton et al., 2002). Men generally have a smaller hip width to femoral length ratio (Horton & Hall, 1989), however, this does not appear to significantly impact the range of motion or position of rotation of the hip that females are capable of when compared to males (Simoneau, Hoenig, Lepley, & Papanek, 1998). Women commonly have a greater range of motion at the hip in general when compared to males, primarily as a result of having greater internal rotation at the hip. The greatest anatomical difference between males and females that contributes to biomechanical running variables is the quadriceps angle, better known as the Q angle (Horton & Hall, 1989), which is the angle created by connecting the line created by the anterior superior iliac spine and the midpoint of the patella to the line created by the midpoint of the patella and the tibial tubercle (Ferber et al., 2003).

Differences in Q angle may be a significant indicator of risk for such injuries as PFPS. A normal Q angle is between 6 and 27 degrees, with the average being approximately 15 degrees (Aglietti, Insall, Walker, & Trent, 1975). A larger Q angle causes the patella to shift laterally, creating a medial tilt and a medial rotation while a smaller Q angle causes the patella to shift

medially, causing a lateral tilt and a lateral rotation (Mizuno et al., 2001). The tracking pattern of the patella is altered by large Q angles, and the result is a lateral shift of the patella which may increase the lateral patellofemoral contact pressure, leading to patellofemoral pain. Additionally, the lateral shift of the patella caused by a greater Q angle could increase the likelihood of subluxation or dislocation of the patella. However, the medial shift of the patella caused by a smaller Q angle may create increased medial tibiofemoral contact pressure.

Biomechanical differences in running technique also exist between men and women. Females generally exhibit increased hip adduction and knee abduction, and absorb more energy at the hip joint during stance (Ferber et al., 2003). Females also tend to have greater hip internal rotation and knee external rotation positions, and absorb more energy at the hip and knee joints while running. Females have been shown to have significantly greater peak hip adduction angles, greater hip negative work in the frontal plane, increased peak hip adduction velocity, and greater peak knee abduction angles. When looking at the running biomechanics in the transverse plane, females usually show much greater peak hip internal rotation angle, hip negative work, and peak hip external rotation velocity (Ferber et al., 2003).

Common overuse injuries in running

Patellofemoral pain syndrome.

The knee is one of the most common injury sites in runners (Lun et al., 2004; Maughan & Miller, 1983; Taunton et al., 2002). Taunton et al found that 42 percent of 2,002 running injuries occurred at the knee, of which anterior knee pain was the most frequent complaint (Thomee, Augustsson, & Karlsson, 1999). A study of injuries among marathon runners also found that 28 percent of the injuries occurred at the anterior knee (Maughan & Miller, 1983). PFPS is one of the most commonly diagnosed injuries in runners (Lun et al., 2004; Taunton et al., 2002) and

anterior knee pain is the primary symptom that individuals with PFPS experience (Thomee et al., 1999). Individuals with PFPS will often experience pain during and after activity, while walking up and down stairs, when squatting, and while sitting with flexed knees (Thomee et al., 1999). Anterior knee pain is often heightened by downhill running and speed training, which increases the damage to the flexor and extensor role of the knee (Rolf, 1995). Ferber and colleagues (2002) established that the kinetic factors most likely to cause knee injuries, including PFPS, are high GRFs, and high knee joint forces and moments (as cited by Messier et al., 2008). As previously mentioned, peak patellofemoral joint compressive forces can be as high as 4 to 7 times body weight during running (William Edwards, 2009; Flynn & Soutas-Little, 1995; Messier et al., 2008), which can lead to overuse injury if appropriate recovery time is not allowed.

The standard alignment of the lower extremity predisposes the patella to lateral forces due to the quadriceps force vector and the patellar tendon force vector being non-collinear (Powers, 2003). Scott and Winter (1990) found the patellofemoral joint force to be as high as 7 times body weight during the stance phase of running due to the need to balance the quadriceps and patellar tendon force vectors, or the Q angle (Ferber et al., 2003; Powers, 2003). A larger Q angle, again, causes the patella to shift laterally which is thought to increase the lateral patellofemoral contact pressure, leading to knee pain (Mizuno et al., 2001). The Q angle can vary based on anatomical structure, however, it can also be affected by abnormal movements of the lower extremity (Powers, 2003). Distally, external rotation of the tibia will increase the Q angle, and proximally, an increased internal rotation of the femur will also increase the Q angle (Powers, 2003; Powers, Ward, Fredericson, Guillet, & Shellock, 2003; Souza & Powers, 2009). Excessive frontal plane motions will also influence the patellofemoral joint. A valgus knee position will increase the Q angle, while a varus position will decrease it. Knee valgus can be

caused by excessive femoral adduction or by excessive tibial abduction, resulting from excessive pronation, or a combination of the two (Powers, 2003). Taunton et al (2002) found that 27 percent of the individuals with PFPS exhibited knee valgus, and 19 percent of the individuals with PFPS showed internal rotation of the femur. Additionally, 6 percent of the individuals with PFPS had high Q angles. Females tend to have greater Q angles than males (Horton & Hall, 1989), which is one reason they tend to develop PFPS more than males.

Iliotibial band friction syndrome.

Iliotibial band friction syndrome is the second leading cause of knee pain in runners behind PFPS, and is the most common cause of lateral knee pain (Taunton et al., 2002). The iliotibial (IT) band originates at the iliac crest and terminates at Gerdy's tubercle and the fibular head, and passes over the lateral femoral condyle. As a result, movements such as increased femoral adduction and internal rotation of the knee leads to increased strain on the IT band (Hamill et al., 2008). In runners with ITBFS, the posterior edge of the IT band impinges against the lateral femoral epicondyle just shortly after initial contact. This impingement occurs at 20 to 30 degrees of knee flexion and repetitive irritation can lead to prolonged inflammation (Orchard, Fricker, Abud, & Mason, 1996). The principal symptom of ITBFS is a sharp or burning pain on the lateral side of the knee. Runners with ITBFS have reported that running downhill, lengthening their stride, and sitting with the knee flexed for prolonged periods of time irritates the condition (Fredericson & Wolf, 2005; Noble, 1980). Running downhill likely causes irritation because it involves landing in a more extended knee position which requires moving through a greater amount of knee flexion during the stance phase (Hamill et al., 2008). In severe cases, the pain may be produced when simply walking or going down stairs (Fredericson & Wolf, 2005).

Females who develop ITBFS and those who have a history of ITBFS have been found to land in greater hip adduction and knee internal rotation and to remain in greater adduction and knee internal rotation throughout stance, and also have a significantly greater peak hip adduction angle (Ferber, Noehren, Hamill, & Davis, 2010; Noehren, Davis, & Hamill, 2007). Individuals with ITBFS also exhibit greater femoral external rotation than healthy individuals (Noehren et al., 2007), and a significantly greater rear foot inversion moment (Ferber et al., 2010). Taunton et al (2002) has also found runners with ITBFS to exhibit a varus knee alignment. Increased internal rotation at the knee increases the load applied to the tissues of the knee joint, including the IT band (Hamill et al., 2008), and increased hip adduction can place greater strain on the IT band. This combination puts the IT band at an increased risk for injury (Ferber et al., 2010). A study comparing female runners who developed ITBFS later on to female runners who did not reported that the ITBFS group displayed a significantly greater strain rate on the IT band in the affected limb compared to both the unaffected limb and the control group (Hamill et al., 2008).

When comparing runners with ITBFS to an uninjured group of runners, Messier et al (1995) found weekly mileage and maximum normalized braking force to be significantly greater in the injured group than the uninjured group. However, it was noted that excessive mileage alone is unlikely to cause an overuse injury unless combined with etiologic risk factors. Though gender was not found to be a discriminator for risk of injury, it was noted that height and strength were found to be advantageous for avoiding injury which was partially explained by gender. Females, therefore, are at a greater risk of sustaining ITBFS. The maximum braking forces in the injured group were significantly attenuated, however, the vertical GRFs and loading rates were lower compared to the control group, which is the opposite of what would be expected. An attenuated braking force may be the result of a lengthened stride, as it has been

reported that shock attenuation increases as stride length increases (Mercer, DeVita, Derrick, & Bates, 2003). This results in more energy being absorbed by the lower extremity joints, particularly the knee joint (Mercer et al., 2003).

Stress fractures.

Distance runners are highly prone to stress fractures. Stress fractures are defined as a partial or complete fracture of the bone resulting from repetitive, submaximal loading (Zadpoor & Nikooyan, 2011). Repetitive loads that cause bending of the tibia will induce a stress injury at the site at which maximum bending occurs (Beck, 1998). Submaximal forces will not result in a stress fracture if there is time for remodeling of the bone as the damage accumulates incrementally (Zadpoor & Nikooyan, 2011). Runners generally miss several weeks of training and competition in order to allow for the bone to heal sufficiently. One study analyzing stress fractures in athletes at a Division I university concluded that the average return to play time for athletes who sustained stress fractures was 8.4 weeks (Arendt, Agel, Heikes, & Griffiths, 2003). Similarly, Beck (1998) determined that the recommended rest period for individuals with tibial stress fractures is 3 to 16 weeks, with the average being 8.3 weeks. This is a considerable amount of time considering the average collegiate athletic season is only about 16 weeks.

Specific activities are more commonly connected with stress fractures than others, with running being the most frequently cited (Reeder, Dick, Atkins, Pribis, & Martinez, 1996). This is the result of numerous cyclical loads, as is experienced with running, causing damage to the bone (Edwards et al., 2009). One of the risk factors likely to be the primary reason for the high stress fracture rate in runners is high GRFs. Running is a highly repetitive activity, and involves very large forces being applied through the lower extremity at a rate of approximately 90 times a minute (Lorimer & Hume, 2014). Vertical GRFs during running can be as high as two to four

times the individual's body weight (Nigg, Cole, & Bruggemann, 1995), which can lead to bone deformation. The higher the GRF, the GRF loading rate, or the number of repetitions, the greater the bone deformation will be (Bennell et al., 2004). Milner, Ferber, Pollard, Hamill & Davis (2006) conducted a study comparing the differences between runners who had sustained tibial stress fractures and those who had never sustained a lower extremity bone injury and found that the vertical GRFs were significantly greater in the injured group compared to the un-injured group. However, in contrast, a study by Bennell et. al. (2004) also compared runners who had sustained a tibial stress fracture in the past four years with those who had never had a stress fracture and found no difference between the GRFs in the two groups. This is an indication that although high GRFs are a likely cause of stress fractures there may be another factor contributing to the strain on the tibia. Therefore, further research is necessary to determine whether GRFs are associated with stress fractures. The rate of force development may also be associated with stress fractures. Burr et al (1996) analyzed strain rates in Israeli infantry recruits and found that tensile and shear strain rates are the greatest during vigorous running activities. The strain gauges were placed at locations in which the recruits most often develop stress fractures, thus they concluded that there is an association between high strain rates and stress fractures.

In a review of 180 cases of stress fractures observed over a two-year period at a sports medicine clinic, the tibia and fibula were among the top five most common locations for a stress fracture (Brukner, Bradshaw, Khan, White, & Crossley, 1996). The most common sport that sustained stress fractures was track and field, followed by distance running. The site that was most commonly injured among the track and field and distance athletes was the tibia. In contrast, Arendt, Agel, Heikies, & Griffiths (2003) examined the physician records of 6000 athletes from a Division I university over a 10-year period and found that stress injuries were relatively

uncommon in the population they chose. However, they found that the population stress injuries to the bone occurred in most often was distance runners, and of these the most common location was the tibia.

Whether males or females are more likely to sustain stress fractures is unclear. Brukner and colleagues (1996) found that stress fractures were more common in males than females (56.7% versus 43.3%). In contrast, Arendt, Agel, Heikes, & Griffiths (2003) revealed that females were almost twice as likely to obtain stress fractures as males (62% versus 38%).

Uphill versus downhill running

Uphill running is often used as a training mechanism to improve running performance and fitness due to the increased energy expenditure that is required as illustrated by an increased heart rate and heightened VO_2 (Cai et al., 2010; Johnny Padulo, Powell, Milia, & Ardigo, 2013; Slawinski et al., 2008). Barnes, Hopkins, McGuigan, and Kilding (2013) carried out a study analyzing five different interval hill training intensities and found that, although no specific hill training method was associated with greater improvements in racing performance, there was an increase in running performance across all five groups. There were improvements made in total five kilometer performance time, peak speed, and aerobic measures such as maximal and submaximal VO_2 , along with stride rate and ground contact time. This shows that incorporating uphill training in running programs can improve both physiological and spatiotemporal parameters relevant to running performance. Inclines are also found in most long distance running events. For example, the Mt. Lemmon Marathon in Arizona has an elevation climb of 5,384 feet, and the Pikes Peak Marathon in Colorado has an elevation climb of 7,901 feet (Maclin, 2016). The majority of shorter running races also have inclines of varying degrees that

must be tackled. Because of the number of endurance races that involve hills, hill training is commonly included in most training programs.

According to the 2015 Annual Report for US Marathons (*2015 Annual report for US marathons*, 2016) there were 540,000 marathon finishers in 2015. The result of this was 14.1 million cumulative racing miles, and approximately 303 million cumulative training miles. The majority of the marathon participants last year were between the ages of 30 and 50, with the 0-29 age group not far behind. Of these racers, 56 percent were men and 44 percent were women. The most popular marathons in the United States last year were the New York City Marathon with almost 50,000 finishers, the Chicago Marathon with 37,500 finishers, the Boston Marathon with 26,600 finishers, and the L.A. Marathon with 21,900 finishers. Many of these courses are known for the steep inclines that occur in the courses. However, along with these inclines come significant declines. Many marathons actually have greater elevation losses than gains (Maclin, 2016). The Boston Marathon has an elevation loss of 1,225 feet, and a net loss of 442 feet. The L.A. Marathon has an elevation loss of 1,270 feet and a net elevation loss of 395 feet. The Big Sur International Marathon, though it did not make the top 10 most popular marathons, has an elevation loss of 2,528 feet and a net elevation loss of 346 feet.

Many distance runners will use hill training as a way to improve their speed on both the uphill and downhill portion of races. Despite the common perception that downhill running is easy and does not need to be practiced, the only way to become better and faster at racing downhill is to train downhill. Paradisis, Bissas, and Cooke (2009) conducted a study analyzing different training methods and determined that when training uphill and downhill for 8 weeks the participants' max running speed increased significantly compared to the group that trained only on a level surface. The step rate was higher, the ground contact time decreased, and the step time

decreased, all of which showed significant differences. These differences illustrate how training on sloping surfaces is more beneficial than training on only a flat surface. The inclusion of hill training in running programs necessitates knowing how uphill and downhill running affects the biomechanics of the individual in order to avoid potential injury to the lower extremity.

Kinetics of hill running.

Running is an activity in which the body experiences repeated impacts with the ground. For example, during a 5 kilometer race, an individual impacts the ground over 3,000 times. These impacts are characterized by a peak GRF, an impact shock caused by rapid deceleration of the lower extremity, and an impact shock wave caused by an initiation of a wave of acceleration and deceleration that is transmitted through the body (Chang & Kram, 1999; Shorten & Winslow, 1992). The GRF is equal to the net force acting on the body's COM and the average acceleration of the whole body (Shorten & Winslow, 1992). During heel-toe running the peak impact force is generally between two and three times body weight (Cavanagh & LaFortune, 1980). Changing foot contact patterns generally causes leg shock to decrease during uphill running, and to increase during downhill running. In addition, frequency and amplitude of impact peaks increase with increased running speed (Shorten & Winslow, 1992).

Ground reaction forces.

During uphill running the vertical GRFs tend to decrease when compared to level and downhill running (Gimenez, Arnal, Samozino, Millet, & Morin, 2014; Gottschall & Kram, 2005). The parallel, or anteroposterior, braking forces are also decreased when running on an incline, while the parallel propulsive forces increase when running uphill (Chang & Kram, 1999; Gottschall & Kram, 2005). Additionally, the normal impact loading rates are generally decreased

when running uphill compared to level or downhill running (Gimenez et al., 2014; Gottschall & Kram, 2005).

In contrast, downhill running tends to produce very different GRFs compared to either level or uphill running. GRFs are a likely contributor to numerous running injuries (Milner et al., 2006). Multiple studies have worked to determine how running downhill changes the impact forces in the vertical direction as well as the braking forces in the horizontal direction. Burr et al. (1996) measured tibial strains in Israeli infantry recruits under various conditions, and found that tensile and shear strains were significantly greater during downhill running compared to level sprinting. The studies by Telhan, et al. (2010), analyzing 21 runners working at a 7% incline and decline in addition to level running; Yokozawa, et al. (2005), looking at 6 male distance runners running at 0%, -3.2%, -6.4%, and -9.1% grades; and Gottschall and Kram (2005), assessing 10 men and women running up and downhill at 0°, 3° (5.2%), 6° (10.5%), and 9° (15.8%) grades, all found that the peak vertical impact forces were significantly greater during downhill running than during level running. The vertical impact loading rate has also been found to be significantly greater in downhill running (Gimenez et al., 2014; Gottschall & Kram, 2005). However, Telhan, et. al. (2010) and Yokozawa, et. al. (2005) found no differences in the parallel braking or propulsive forces, while Gottschall and Kram (2005) and Chang and Kram (1999) found there to be significantly greater parallel braking forces and significantly smaller propulsive forces in downhill running in comparison to level running.

Joint Kinetics.

When analyzing power output, energy absorption, and work performed by the lower extremity joints the knee performs significantly greater amounts of negative work when running downhill at a -4% grade than when running on a level surface (Buczek & Cavanagh, 1990). The

percent of time in which the knee was performing negative work was greatly increased at -4% as well. This may be the reason for the increased soreness felt by runners when training downhill as the knee must absorb more energy running downhill compared to when running on a level surface (Telhan et al., 2010; Yokozawa et al., 2005).

Peak power absorption at the ankle showed the greatest difference between level running and downhill running at a -8.3% grade; it was nearly 1.5 times greater in downhill running (Buczek & Cavanagh, 1990). In addition, the amount of negative work performed by the ankle was doubled in downhill running, and the amount of time spent performing negative work was greatly increased as well. Buczek and Cavanagh (1990) concluded that the increased muscle soreness at the ankle is likely a result of the significantly greater power absorption and negative work that occurs at the ankle. This is potentially due to the fact that forces are transferred from the foot up through the body, therefore, the ankle must dissipate much larger forces in comparison to the rest of the body (Zhang, Bates, & Dufek, 2000). In contrast, Yokozawa et al. (2005) and Telhan et al. (2010) found no significant differences in peak power absorption at the ankle, but greater power absorption at the hip during downhill running.

When examining differences in joint forces between running downhill at -8.3% and level running were analyzed, researchers found that there were no significant differences in the peak extensor moment at the knee (Buczek & Cavanagh, 1990; Telhan et al., 2010). This was unexpected because the muscle soreness often reported by runners is commonly in the quadriceps, or knee extensor, muscles. Yokozawa, et al. (2005) established that the knee extensor moment does, however, tend to increase more rapidly when running at declines of -6.4% and -9.1% than during level running. They also found the peak plantar flexor moment at the ankle and the hip extensor moment to be smaller during the stance phase when running at the steeper

grades than level running. In contrast, Telhan et al. (2010) determined that there were no significant differences in the lower extremity joint moments when running downhill at a 4% decline at a constant velocity. Additionally, there were no significant differences in the peak joint moments found during the recovery phase of running at any grade (Yokozawa et al., 2005).

Kinematics of hill running.

The kinematics utilized in running can affect the impact shock felt during running. One kinematic change that has the potential to reduce the GRF is increased knee flexion, as well as an increased rate of knee flexion (Derrick, 2004; Frederick, 1986). Foot contact patterns and footwear can also affect the impact force (Frederick, 1986); however, these differences will be neglected for the purpose of this study.

When sprinting uphill significantly smaller trunk and shank angles (measured against the horizontal) have been seen (Paradisis & Cooke, 2001). These changes are thought to contribute to the shorter stride length seen in uphill running. Swanson and Caldwell (2000) found greater hip, knee, and ankle joint flexion at initial contact. Compared to level, the hip went through extension during stance in the incline running condition compared to flexion, range of motion (ROM) at the knee was seen to be significantly smaller during stance, and the ankle was in greater dorsiflexion at initial contact and went through less dorsiflexion motion during incline running. Slawinski et al. (2008) found greater lower extremity flexion during the stance phase of incline running compared to level running. He also found the ROM at the knee and ankle to be smaller when running uphill than when running on a level surface.

In comparison to level and uphill running, the kinematics of downhill running are significantly different. Running downhill has been shown to result in a greater angle at the shank, greater knee extension, and less hip flexion at initial contact (Paradisis & Cooke, 2001). Data

collected on 14 males running at 0% and -4% grades (Mizrahi, Verbitsky, & Isakov, 2000), and on 7 males running at 0% and -8.3% grades (Buczek & Cavanagh, 1990) showed that when running downhill there was significantly less knee flexion at initial contact, but a greater peak flexion angle, and a greater time to peak flexion, as well as a much greater percent of stance in which the flexion velocity was at its peak. This is similar to what Yokozawa, Fujii and Ae (2005) found in their analysis of 6 male distance runners on grades of 0%, -3.2%, -6.4%, and -9.1%. They found that running at steeper declines produced higher peak flexion velocities in the support phase in comparison to running at level and -3.2% grade. These results are thought to be the result of needing to extend further to reach for the lower ground.

The study conducted by Buczek and Cavanagh (1990) also revealed that when running downhill the knee goes through a greater ROM due to the need to cushion the landing. When running downhill at a -8.3% decline the percent of time in which the ankle was in maximum dorsiflexion and was at maximum dorsiflexion velocity were both significantly increased (Buczek & Cavanagh, 1990). Additionally, it was found in a study analyzing the running kinematics of six male distance runners at various speeds on 0%, -3.2%, -6.4%, and -9.1% grades that at initial contact there was more hip extension in downhill running at the -9.1% grade when compared to level running (Yokozawa et al., 2005). When looking at the angle between the lower leg and the treadmill it was found that the leg-treadmill angle increased between running uphill to running downhill, indicating that runners tend to lean forward more when running uphill and to lean backward when running downhill (Gimenez et al., 2014).

Spatiotemporal parameters of hill running.

Variables such as stride length and stride rate are often different during uphill and downhill running than when running on a level surface. However, these parameters can affect

both the kinematics and the kinetics of running. There is an inverse relationship between stride rate and stride length, horizontal distance between the center of mass (COM) and the heel at initial contact, and braking impulse (Heiderscheit, Chumanov, Michalski, Wille, & Ryan, 2011). As stride rate increases, the stride length shortens, and the heel is placed closer to the COM, which reduces the braking impulse. There is also less energy absorbed at the knee and hip when the stride rate is increased, and there is increased energy absorbed at all three lower extremity joints when the stride rate is decreased (Heiderscheit et al., 2011).

Previous research has found that when running on an incline the stride rate is quicker than when running downhill or on a horizontal surface (Gimenez et al., 2014; Swanson & Caldwell, 2000). Additionally, there has been research showing that when running uphill there is a greater percent of time spent in stance compared to level running (Slawinski et al., 2008; Swanson & Caldwell, 2000). There is also a much lower velocity when running uphill, which has been associated with the lower stride rate and stride length in incline running (Slawinski et al., 2008).

A study that compared running at a two percent grade with a manipulated step frequency to a control group with a freely chosen step frequency resulted in significant differences in contact time, flight time, step frequency, and step length (J Padulo et al., 2012). The group with a manipulated step frequency had an average stride rate of 2.93 Hz, while the freely chosen step frequency group had a stride rate of 3.08 Hz. The manipulated step frequency group showed a significantly shorter contact time, longer flight time, and longer step length, all of which are beneficial to running performance. Additionally, a study examining the differences between a combined uphill and downhill training group, a horizontal training group, and an untrained control group determined that the combined hill training showed the greatest improvements

(Paradisis et al., 2009). This was seen in the improved maximum running speed, increased stride rate, decreased contact time, and decreased step time in the combined hill training group, all of which occurred with an unaltered stride length. In contrast, Paradisis and Cooke (2001) examined spatiotemporal differences in uphill, downhill, and horizontal running and found no significant differences in stride rate, contact time, or flight time for any condition. However, they did find that uphill running showed a significantly shorter stride length and a significantly shorter flight distance compared to horizontal and downhill running. Similarly, Slawinski et al. (2008) found no differences in flight time between level and incline running, Gottschall and Kram (2005) found stride kinematics to be very similar between uphill and downhill running except for an increased stride rate when running uphill, and Paradisis and Cooke (2001) found that stride length was significantly greater in downhill running.

The study by Heiderscheit and colleagues (2011) revealed that when the stride rate is increased by 10 percent from the preferred stride rate there is greater knee flexion at initial contact, a smaller peak knee flexion angle during stance, and smaller peak flexion and adduction angles with smaller internal rotation and peak abduction moments at the hip. These changes are likely a result of the decreased stride length and altered lower extremity posture that occur when the stride rate increases. Excessive hip motion during running has been associated with anterior knee pain and iliotibial band syndrome (Ferber et al., 2010; Noehren et al., 2007), therefore, an increased stride rate, and thus a decreased stride length, would be beneficial for reducing the risk of injury.

Yokozawa, et. al. (2005) assessed the spatiotemporal parameters of 6 male distance runners at 0%, -3.2%, -6.4%, and -9.1% grades at various running speeds and found that there were no significant differences in running velocity, stride length, or stride frequency among the

grades or the running speeds. However, they did find that there was more time spent in the stance phase at the -9.1% grade compared to the -3.2% grade. Similarly, Chang and Kram (1999) and Gimenez, Arnal, Samonizo, Millet, & Morin (2014) both analyzed differences in running when an aiding and an impeding horizontal force was applied in order to simulate uphill and downhill running. Chang and Kram found no significant changes in stride rate, but did find significant changes in ground contact time. In contrast, Gimenez et al. found that stride rate decreased significantly from level running to downhill running. A decrease in stride rate and an increase in stride length during downhill running in comparison to level running may put distance runners at an increased risk for lower extremity injury.

Muscle activity in hill running.

The amount of work required by the muscles during uphill running is different from that of level running. Sloniger, Cureton, Prior, and Evans (1997) found that during uphill running the most activated muscles were the adductors, biceps femoris, gluteal muscles, gastrocnemius, and the vastus muscle group, while the least activated were the rectus femoris, soleus, tibialis anterior, and gracilis. However, it was determined that even the most active muscles were not fully activated during uphill running. Padulo, Powell, Milia, and Ardigo (2013) found decreased muscle activity in the tibialis anterior, rectus femoris, vastus medialis, and medial gastrocnemius, and increased activity in the gluteal major with increasing grade. This is similar to what Sloniger et al. found in his study, however, Sloniger and colleagues found the vastus group and the medial gastrocnemius to be more active in incline running. Cai et al. (2010) determined that rectus femoris and gastrocnemius muscle activity was greater during uphill running compared to horizontal running. This is contradictory to what Sloniger et al. and Padulo et al. found. Cai and colleagues found no differences in the tibialis anterior and soleus muscle activity between uphill

and horizontal running. Cai et al. reasoned that the increased activity in the rectus femoris was likely a sign that running uphill requires greater knee flexion, knee extension, and hip extension in order to achieve the climbing movement. In addition, they proposed that the increased gastrocnemius muscle activity was due to an increased propulsive force. During the stance phase of uphill running Swanson and Caldwell (2000) found large increases in muscle activity in the gastrocnemius, soleus, rectus femoris, vastus lateralis, and gluteus maximus, but lower amplitudes in the medial hamstring and biceps femoris. Slawinski et al. (2008) found lower muscle activation in the biceps femoris and semitendinosis during the stance phase of uphill running than level running, but found no significant differences in muscle activation between uphill and level running during the braking phase.

The muscle groups that seem to be most affected by downhill running are the gluteal muscles, the quadriceps, and the calf muscles, all of which are the extensor muscles of the hip, knee, and ankle (Buczek & Cavanagh, 1990). It would be expected that the extensor muscles would show a greater amount of muscle activation when running downhill because of the braking action needed to prevent oneself from falling down the hill. Cai et al. (2010) studied the effect of uphill, level, and downhill running on muscle activity and delayed onset muscle soreness (DOMS) in 8 active males running at 55% of their maximum VO_2 . Electrode placement was limited to five major muscles in order to eliminate potential noise. The muscles analyzed were the rectus femoris, biceps femoris, tibialis anterior, lateral gastrocnemius, and the soleus. Muscle soreness set in immediately after each running trial, and DOMS was felt at 24 hours. It was determined that downhill running induced the greatest DOMS. Additionally, running downhill at only 64% of heart rate max produced a greater DOMS effect than running uphill at 90% of heart rate max. This is a strong indicator that eccentric muscle activity plays a much

larger part in the onset of DOMS compared to exercise intensity (Cai et al., 2010). The gastrocnemius, soleus, and tibialis anterior all had greater muscle activity in downhill running than level running, however, only the gastrocnemius showed a significant difference between the two conditions. The biceps femoris and rectus femoris both showed more muscle activity in level running when compared to downhill running. It is thought that the easier physiological workload required in downhill running is the reason for the smaller amount of rectus femoris activity in downhill running (Cai et al., 2010).

During downhill running, the muscles contract eccentrically (Brown et al., 1999; Cai et al., 2010). Eccentric muscle activity often results in a greater force being applied to fewer motor units (Cai et al., 2010). This pattern of muscle contraction may be the reason for the increased DOMS linked to downhill running. An increase in muscle activity in the extensor muscles when running downhill would be expected. However, the use of fewer motor units in eccentric muscle contractions may result in decreased muscle activity, as was seen by Cai et al (2010). Additionally, there is evidence of collagen breakdown in the muscles caused by eccentric muscle activity (Brown et al., 1999). The breakdown of collagen during eccentric muscle contractions may potentially place individuals at a greater risk for further connective tissue injury (Brown et al., 1999).

Conclusion

Running downhill produces biomechanics than running on level or uphill surfaces (Buczek & Cavanagh, 1990; Cai et al., 2010; Gottschall & Kram, 2005; Mizrahi et al., 2001; Yokozawa et al., 2005). Downhill running shows differences that may increase an individual's risk for injury in the long term as it exhibits several factors that have been found in individuals who have experienced running overuse injuries. However, the majority of the prior research done

on this topic has primarily looked only at males. It is important to understand the impact of downhill running on females as well in order to maximize training and avoid injury. Additionally, most researchers have looked at grades that are the equivalent of rolling hills. Most distance runners will run down much steeper grades than the ones that have been studied in the past. We do not know whether the risks associated with downhill running plateau at a certain grade, or whether the risks continue to become more severe with increasing declines. Therefore, the purpose of this study is to compare the kinematics, kinetics, spatiotemporal parameters and muscle activity on different downhill grades in female distance runners.

Chapter 3

Methodology

Participants

A total of 15 female distance runners participated in this study (Age: 23.5 ± 4.9 years, Height: $1.7 \pm .06$ m, Weight: 57.8 ± 6.8 kg). The participants were regular runners who trained 35 ± 13 miles a week. Participants were in good cardiovascular health, had no history of any lower extremity injury resulting in an inability to run for at least the six months prior to the study, and were free of any chronic musculoskeletal conditions, or neurological problems that would inhibit their ability to run normally.

Participants were recruited from the population of a Division I university and the surrounding areas. Recruitment was accomplished through emails, fliers on campus, and word of mouth. The Institutional Review Board at Ball State University approved the study, and informed consent was obtained from each individual prior to data collection.

Instruments

Motion capture.

Kinematic data collection and spatiotemporal parameters were collected using 15 VICON infrared cameras (Oxford Metrics, Oxford, UK) sampling at 200 Hz. Retroreflective markers were placed on anatomical landmarks following a modified Plug-in-Gait reflective marker set, as

described below. All cameras were calibrated prior to each testing session in order to ensure accurate data collection.

VICON Nexus 2.5 software (Oxford Metrics, Oxford, UK) was used during the running trials for data collection. All cameras and the AMTI force plates were connected to VICON Nexus, and the software was used to adjust the parameters, calibrate the cameras, and record the running trials.

Force-instrumented treadmill.

All running trials took place on an AMTI force-instrumented treadmill (Advanced Mechanical Technology, Inc., Watertown, MA, USA) comprised of two tandem belts; each with a force plate beneath it. Only the first, or front, force plate was used for data collection. The force plate was used to collect GRFs and to calculate joint moments at all three lower extremity joints. Forces and moments were measured in the x, y, and z directions. The force plates were connected to MSA-6 amplifiers (Advanced Mechanical Technology, Inc., Watertown, MA, USA), sampling at a rate of 2000 Hz. The treadmill was equipped with handrails on either side of the belts, and emergency stop buttons were located on one of the railings and by the collection computer. The speed and grade of the treadmill were controlled using an End-to-End software (Advanced Mechanical Technology, Inc., Watertown, MA, USA).

Electromyography.

Muscle activity of the vastus medialis, gluteus medius, biceps femoris, medial gastrocnemius, and tibialis anterior on the dominant limb was collected during each running trial using five Delsys Trigno wireless electromyography (EMG) sensors (Delsys Inc., Boston, MA, USA). EMG data was collected at a sampling rate of 2000 Hz. The raw EMG signals were band-pass filtered at 20-500 Hz, and the root mean square amplitude of each signal was then

calculated using a 50ms window. EMG activity of each muscle was normalized to the maximum voluntary contractions for each participant (Ball & Scurr, 2013).

Cybex dynamometer.

Maximum Voluntary Isometric Contractions (MVIC) at the hip, knee, and ankle were collected using a Cybex norm dynamometer (Cybex International, NY, USA) sampling at a rate of 100 Hz and a MicroFet2 manual muscle tester (Hoggan Health Industries, UT, USA). The MVICs collected were abduction of the hip, flexion and extension of the knee, and dorsiflexion and plantarflexion of the ankle. The Cybex was equipped with handles to hold onto, and a seatbelt.

VICON Nexus 2.5 software (Oxford Metrics, Oxford, UK) was used to collect the muscle activity during the MVICs.

Experimental Protocol

All data collection took place in the Biomechanics Laboratory at Ball State University. All of the instruments and equipment were calibrated prior to participant arrival in order to ensure accurate data.

Participant preparation.

Upon arrival, the participant filled out a health history and running questionnaire, as well as an ACSM health questionnaire (Appendix A) in order to determine eligibility based on the inclusion and exclusion criteria established, and signed an informed consent. They were then provided compression clothing and standardized Nike shoes (NIKE, Inc., Portland, OR, USA) to change into. Anthropometric measurements were taken, which included: height in millimeters, mass in kilograms, and ankle width, knee width, inter-ASIS distance, and leg length in millimeters. Wireless EMG sensors were then placed on the belly of the gluteus medius, vastus

medialis, biceps femoris, medial gastrocnemius, and tibialis anterior on the dominant limb. Two accelerometers were also placed on the proximal and distal end of the tibia on the dominant limb. Prior to EMG and accelerometer placement the sites were shaved, scrubbed with sandpaper, and cleaned with alcohol in order to reduce impedance and to ensure a good transmittance of the signals. Following EMG placement, retroreflective markers used for the motion capture were placed on bony landmarks on the lower extremity and the trunk using double-sided tape and Tuf-Skin (Cramer Products Inc., Gardner, KS, USA). The marker placement followed a modified Plug-in-Gait marker set with individual markers placed on the right and left shoes between the first and second toe, the calcaneus, and the head of the fifth metatarsal, the right and left medial and lateral malleoli, the right and left medial and lateral femoral condyles, the right and left posterior superior iliac spine, the sternum, the jugular notch between the clavicles, and on the right and left acromion processes of the shoulders. Clusters of reflective markers were also placed on the lateral aspect of the right and left mid shanks, the lateral aspect of the right and left mid thighs, and the anterior aspect of the right and left iliac crests. The EMG sensors and the cluster sets were further secured with PowerFlex (Andover Healthcare, Inc., Salisbury, MA, USA) cohesive bandage to ensure they did not fall off during the running trials. This marker set defined the 3-dimensional kinematics of the trunk, pelvis, left and right thighs, left and right shanks, and both feet. Following reflective marker and EMG placement, the participant was moved into the data collection area in order to ensure all of the markers were visible and that the EMG and accelerometer sensors were transmitting clear signals.

Data collection.

Prior to the running trials, a static-subject calibration and ROM trials were collected by the VICON motion capture system. The static calibration consisted of standing on the treadmill

with the feet shoulder width apart, and the arms reaching out to the sides with the thumbs pointing down. Functional joint ROM trials of the right and left hip and knee were then collected. Hip ROM consisted of moving the leg in a five-point star pattern followed by one circle. Knee ROM consisted of extending the hip back, and then flexing and extending the knee through 20-30 degrees of motion five times. The ROM trials were used to construct functional joints during data processing. Following the functional joint ROM trials, the medial knee and ankle markers were removed and the participants warmed up at a self-selected pace for 5 minutes on a 0% grade on the AMTI Force Instrumented Treadmill. Since participants were not able to adjust the speed themselves, as the speed was controlled by the End-to-End software on the computer, the treadmill was started at 3.0 m/s and then adjusted according to the participants' request. Following the warm-up, the participants were given a short rest while they were fit with a harness that was attached to the ceiling for safety. The harness was used as a precaution should any of the participants' slip, and to ensure the participants were comfortable running with a normal stride on the steepest declines. The order of the running trials was randomized using a random number generator in order to avoid the effect of potential fatigue and to eliminate any potential learning effect. The participants ran on the AMTI force instrumented treadmill for a total of 5 running conditions. The grades of the treadmill were 0%, -5%, -10%, -15%, and -20% and the speed of the treadmill was set at a constant speed of 4.0 meters per second. The participants were instructed to step to either side of the treadmill belt while the treadmill was started and the force plate was zeroed, and then to step on and run. Each trial lasted approximately 2 minutes; long enough for the participant to adopt a comfortable and consistent running stride and for four, good 7-second trials of data to be collected. A trial was considered good if all reflective markers were visible, and the EMG sensors were picking up the muscle

activity appropriately. The participants were instructed to run on the first, or front, belt of the treadmill. The participants were given 5 minutes of rest between each trial to avoid becoming fatigued during the data collection. Following the running trials, the participants completed MVICs using a Cybex dynamometer and a manual muscle tester. MVICs for knee flexion at thirty degrees and extension at sixty degrees from the anatomical zero, ankle dorsiflexion and plantarflexion at fifteen degrees, and hip abduction were collected on the dominant limb. The trials were 3 seconds long, and the participants were given a minute of rest between each trial. The EMG data from the MVICs was used to normalize the EMG data from the running trials.

The entire procedure took between 1.5 and 2 hours to complete for each participant from the time the participant arrived at the lab to the time they left.

Data processing.

VICON Nexus 2.5 (Oxford Metrics, Oxford, UK) was used to reconstruct and label the markers of the static calibration, functional joint ROM trials, and the running trials. All of the data was then transferred over to Visual 3D v.5 (C-Motion, Inc., Fermantown, MD, USA) where a model was created using the functional joint ROM data. Matlab R2016 and Visual 3D software was used to calibrate the force plates at the different declines. Force plate data was manually zeroed in Visual 3D. A recalculation pipeline was run to calculate the lower extremity joint angles, moments, and powers, and an automatic gait events pipeline was run to calculate spatiotemporal parameters. Dependent variables were stride rate and step length; peak vertical active GRF, peak braking force, peak propulsive force, peak and average loading rate; hip, knee, and ankle joint moments, angles, and powers in the frontal, sagittal, and transverse planes; and muscle activity. Variables were averaged from 5 foot strikes across the three best trials.

Statistical analysis.

Repeated measure, multivariate analysis of variances (MANOVA) were run to test for significant differences between the 5 running conditions in order to answer the research questions. If the main MANOVA showed significant differences, the univariate analysis of variance (ANOVA) results were then examined for significance. If the univariate tests showed significance, pairwise comparisons with a Bonferroni correction factor were further utilized to determine which conditions the significant differences occurred between. The significance was set at $p < 0.05$, and IBM SPSS v.24 (SPSS Inc., Chicago, IL, USA) was used to run the statistical tests.

Chapter 4

Results

The results chapter is divided into multiple sections. Ground reaction forces will be discussed first, followed by joint kinetics, joint kinematics, spatiotemporal parameters, and muscle activity. The joint kinetics section is further divided into joint powers, and joint moments in each of the three planes of motion. The joint kinematics section is also divided into each plane of motion.

Ground Reaction Forces

The repeated measure MANOVA was significant for peak vertical GRF ($p < .001$), peak propulsive force ($p < .001$), and for the average ($p < .001$) and instantaneous loading rates ($p < .001$), while it was not significant for the peak braking force ($p > .05$). Pairwise comparisons further revealed significant differences between each of the grades. The peak vertical GRF increased from 0% to -15% ($p = .014$) and from 0% to -20% ($p < .001$) (Figure 1). The -15% grade was also greater than -5% ($p = .003$) and -10% ($p < .001$), and -20% was greater than each of the other grades ($p < .001$). The peak propulsive force was significantly smaller at -20% compared to 0% ($p = .021$). The average loading rate and the instantaneous loading rate were both greater in each of the downhill grades compared to level. The average loading rate also increased in a dose-response relationship, however, 0% and -5% were not significantly different ($p > .05$). The -10%

grade was greater than 0% and -5% ($p=.004$), -15% was greater than -10% ($p=.025$), and -20% was greater than -15% ($p=.013$). The instantaneous loading rate increased in a dose-response relationship by 10% increments from 0% to -20% (Figure 2). The -10% grade was greater than 0% ($p=.008$), -15% was greater than -5% ($p=.035$), and -20% was greater than -10% ($p=.037$) (Table 1).

Joint Kinetics

Power absorption and production at the lower extremity joints.

The repeated measure MANOVA was significant for peak ankle power absorption ($p=.002$), peak ankle power propulsion ($p<.001$), peak knee power absorption ($p<.001$), peak knee power propulsion ($p<.001$), and peak hip power absorption ($p<.001$), but was not significant for peak hip power propulsion ($p>.05$). Pairwise comparisons further revealed that peak ankle power absorption increased from 0% to -5% ($p<.001$), 0% to -10% ($p=.003$), 0% to -15% ($p=.001$), and 0% to -20% ($p=.017$). Peak ankle power propulsion decreased in each of the downhill grades (except -5%) compared to level, and there was a dose-response relationship (Figure 3). The -10% grade was smaller than 0% and -5% ($p=.001$), -15% was smaller than -10% ($p<.001$), and -20% was smaller than -15% ($p<.001$). Peak knee power absorption was greater in each of the downhill grades (except -5%) compared to level (Figure 3). There was also a dose-response in 10% increments, so -10% was greater than 0% ($p=.001$), -15% was greater than -5% ($p<.001$), and -20% was greater than -10% ($p<.001$). Peak knee power propulsion decreased in each of the downhill grades compared to level, and there was a dose-response relationship, so -5% was smaller than 0% ($p<.001$), -10% was smaller than -5% ($p=.02$), -15% was smaller than -10% ($p=.01$), and -20% was smaller than -15% ($p<.001$). Peak hip power absorption increased significantly from 0% to -20% ($p=.044$) (Table 2).

Sagittal plane joint moments.

The repeated measure MANOVA was significant for the peak dorsiflexion moment ($p<.001$), peak plantarflexion moment ($p<.001$), peak knee flexion moment ($p<.001$), peak knee extension moment ($p=.002$), peak hip flexion moment ($p=.027$), and the peak hip extension moment ($p<.001$). Pairwise comparisons did not show differences in the peak dorsiflexion moment between the downhill grades and level. The peak plantarflexion moment increased from 0% to -5% ($p<.001$), and from 0% to -10% ($p=.043$), but it decreased from 0% to -20% ($p<.001$). Pairwise comparisons also did not show differences between the downhill grades and level in the peak knee flexion moment. The peak knee extension moment decreased from 0% to -5% ($p=.002$). The peak hip flexion moment decreased from 0% to -5% ($p=.001$). The peak hip extension moment was greater in each of the downhill grades compared to level ($p<.05$) (Table 3).

Frontal plane joint moments.

The repeated measure MANOVA was significant for the peak ankle inversion moment ($p<.001$), peak knee adduction moment ($p<.001$), peak knee abduction moment ($p<.001$), and the peak hip adduction moment, but it was not significant for the peak ankle eversion moment ($p>.05$), or the peak hip abduction moment ($p>.05$). The peak ankle inversion moment was greater in each of the downhill grades compared to level ($p<.001$). In addition, -10% and -15% were both greater than -5%. The peak knee adduction moment was greater in each of the downhill grades compared to level, and there was a dose response relationship, but -10% and -15% were not significantly different. The -5% grade was greater than 0% ($p=.004$), -10% was greater than -5% ($p=.004$), and -20% was greater than -10% ($p=.019$) and -15% ($p=.005$). The peak knee abduction moment decreased from 0% to -10% ($p=.001$), 0% to -15% ($p=.010$), and

0% to -20% ($p=.005$). The peak hip adduction moment increased from 0% to -15% ($p=.036$), and 0% to -20% ($p=.042$) (Table 4).

Transverse plane joint moments.

The repeated measure MANOVA was significant for the peak ankle internal rotation moment ($p<.001$), and the peak knee internal rotation moment ($p=.002$), but it was not significant for the peak ankle external rotation moment ($p>.05$), the peak knee external rotation moment ($p>.05$), or the peak hip internal and external rotation moments ($p>.05$). The peak ankle internal rotation moment increased from 0% to -10% ($p=.003$), 0% to -15% ($p<.001$), and 0% to -20% ($p=.001$). In addition, -10% was greater than -5% ($p=.009$), and -15% was greater than -10% ($p=.007$). Peak knee internal rotation was significantly greater in the -15% condition compared to 0% ($p=.011$) and -5% ($p=.002$) (Table 5).

Joint Kinematics

Sagittal plane joint angles.

The repeated measure MANOVA was significant for the peak ankle dorsiflexion angle ($p<.001$), peak ankle plantarflexion angle ($p<.001$), knee flexion angle at initial contact ($p<.001$), peak knee flexion angle ($p<.001$), hip flexion angle at initial contact ($p<.001$), peak hip flexion angle ($p<.001$), peak hip extension angle ($p<.001$), and the trunk extension angle at initial contact ($p<.001$), but it was not significant for the ankle angle at initial contact ($p>.05$). The peak ankle dorsiflexion angle decreased in each of the downhill grades compared to level, and there was a dose-response relationship, but -10% and -15% were not significantly different from each other. The -5% grade was smaller than 0% ($p<.001$), -10% was smaller -5% ($p=.001$), and -20% was smaller than -15% ($p<.001$) and -10% ($p<.001$). The peak plantarflexion angle decreased in each of the downhill grades, and there was a dose-response pattern, however, -15% and -20%

were not significantly different from each other. The -5% grade was smaller than 0% ($p=.008$), -10% was smaller than -5% ($p=.001$), and -15% was smaller than -10% ($p=.001$). The significantly smaller dorsiflexion and plantarflexion angles that resulted from the increasing downhill grade resulted in a decreased ROM at the ankle (Figure 4).

The knee flexion angle at initial contact decreased (increased extension) between level and each of the downhill grades ($p<.001$). The peak knee flexion angle increased from 0% to -20% ($p=.013$). The increased knee extension at initial contact and the increased peak knee flexion resulted in a greater ROM at the knee joint in the downhill conditions (Figure 4). The hip flexion angle at initial contact decreased from level to each of the downhill grades ($p<.001$). The peak hip flexion angle decreased from 0% to -5% ($p=.002$), 0% to -10% ($p=.005$), 0% to -15% ($p=.015$), and from 0% to -20% ($p=.024$). The peak hip extension angle decreased in each of the downhill grades (except -5%) compared to level, and there was a dose-response relationship. The -10% grade had a smaller peak hip extension angle than both 0% ($p=.033$) and -5% ($p=.026$), -15% was smaller than -10% ($p=.010$), and -20% was smaller than -15% ($p=.014$). The decreased flexion at initial contact and the decreased peak extension resulted in a smaller ROM at the hip during each of the downhill grades compared to the level condition (Figure 4). The trunk extension at initial contact increased from level to each of the downhill grades ($p<.001$) (Table 6).

Frontal plane joint angles.

The repeated measure MANOVA was significant for the peak ankle eversion angle ($p<.001$), the hip adduction angle at initial contact ($p<.001$), the peak hip adduction angle ($p<.001$), and the peak hip abduction angle ($p=.004$), but it was not significant for the ankle inversion angle at initial contact, the peak ankle inversion angle, the knee abduction angle at

initial contact, or the peak knee adduction or abduction angles ($p>.05$). The peak ankle eversion angle decreased from 0% to -15% ($p=.001$), and 0% to -20% ($p<.001$). In addition, peak ankle eversion was smaller in the -20% condition than each of the other grades ($p<.01$). The hip adduction angle at initial contact decreased in each of the downhill grades compared to level, and there was a dose-response relationship. The -5% grade had a smaller adduction angle than 0% ($p<.001$), -10% was smaller than -5% ($p<.001$), -15% was smaller than -10% ($p<.001$), and -20% was smaller than -15% ($p<.001$). The peak hip adduction angle also decreased in each of the downhill grades, and there was a dose-response relationship seen, however, -10% and -15% were not significantly different from each other. The angle decreased from 0% to -5% ($p=.034$), -5% to -10% ($p=.034$), -10% to -20% ($p<.001$), and from -15% to -20% ($p<.001$). The hip never became abducted, but the minimum adduction angle increased from 0% to -10% ($p=.007$) and from 0% to -15% ($p=.007$) (Table 7).

Transverse plane joint angles.

The repeated measure MANOVA was significant for the peak ankle external rotation angle ($p<.001$), the peak hip external rotation angle ($p<.001$), and the peak hip internal rotation angle ($p<.001$), but it was not significant for the peak ankle internal rotation angle ($p>.05$), the peak knee internal rotation angle ($p>.05$), or the peak knee external rotation angle ($p>.05$). The peak ankle external rotation angle increased from 0% to -5% ($p=.030$), 0% to -10% ($p=.024$), 0% to -15% ($p=.003$), and from 0% to -20% ($p=.002$). The peak hip external rotation angle increased from 0% to -20% ($p=.002$). from -5% to -20% ($p<.001$), from -10% to -20% ($p=.001$), and from -15% to -20% ($p=.003$). The peak hip internal rotation angle decreased from 0% to -15% ($p=.025$), and from 0% to -20% ($p<.001$) (Table 8).

Spatiotemporal Parameters

The repeated measure MANOVA was not significant for the stride rate or for the step length ($p > .05$) (Table 9).

Electromyography

The repeated measure MANOVA was significant for the average medial gastrocnemius muscle activity ($p < .001$), and for the peak medial gastrocnemius muscle activity ($p < .001$), but it was not significant for the average of peak muscle activity in the gluteus medius, vastus medialis, biceps femoris, or tibialis anterior muscles ($p > .05$). The average medial gastrocnemius muscle activity decreased from 0% to -10% ($p = .012$), 0% to -15% ($p = .017$), and 0% to -20% ($p = .024$). The peak medial gastrocnemius muscle activity increased from 0% to -5% ($p = .003$), but decreased from -5% to -15% ($p = .001$), and from -5% to -20% ($p = .003$) (Table 10).

Tables and Figures

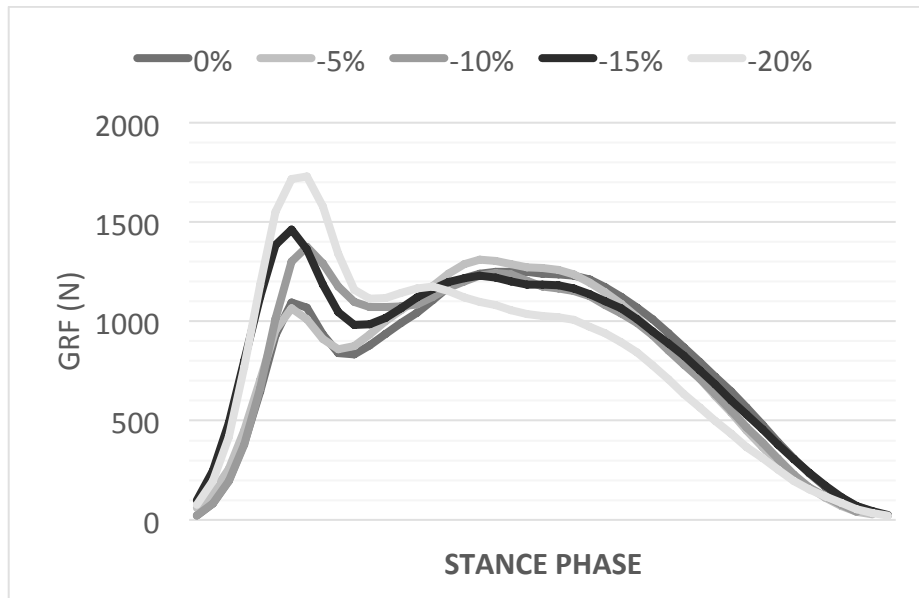


Figure 1: Representation of the peak vertical ground reaction force (N) during the stance phase for each grade.

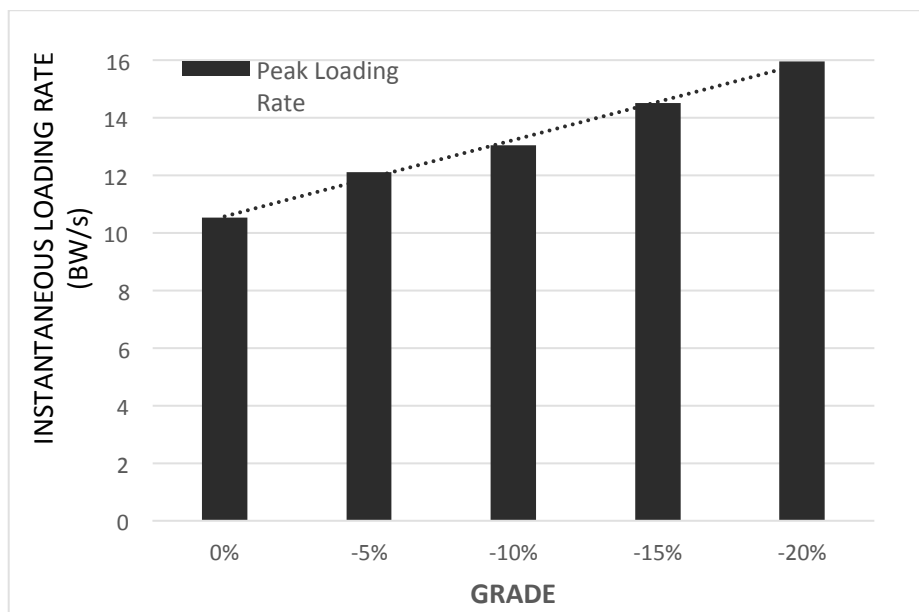


Figure 2: Mean instantaneous loading rate ($\text{BW} \cdot \text{s}^{-1}$) averaged across all subjects for each grade.

Table 1: Mean \pm SD ground reaction forces averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Vertical GRF (BW)	2.48 \pm 0.16	2.47 \pm 0.16	2.53 \pm 0.20	2.76* \pm 0.24	3.01* \pm 0.31
Braking Force (BW)	-0.41 \pm 0.05	-0.42 \pm 0.06	-0.42 \pm 0.06	-0.45 \pm 0.10	-0.43 \pm 0.09
Propulsive Force (BW)	0.38 \pm 0.05	0.39 \pm 0.05	0.38 \pm 0.04	0.36 \pm 0.05	0.33* \pm 0.05
Avg. Loading Rate (BW*s⁻¹)	2.45 \pm 0.37	3.00 \pm 1.15	5.65* \pm 2.46	7.69* \pm 2.20	8.74* \pm 2.17
Peak Loading Rate (BW*s⁻¹)	10.52 \pm 2.53	12.11 \pm 2.64	13.04* \pm 3.63	14.51* \pm 5.16	15.95* \pm 5.21

(*) Indicates a significant difference from the 0% grade ($p < .05$)

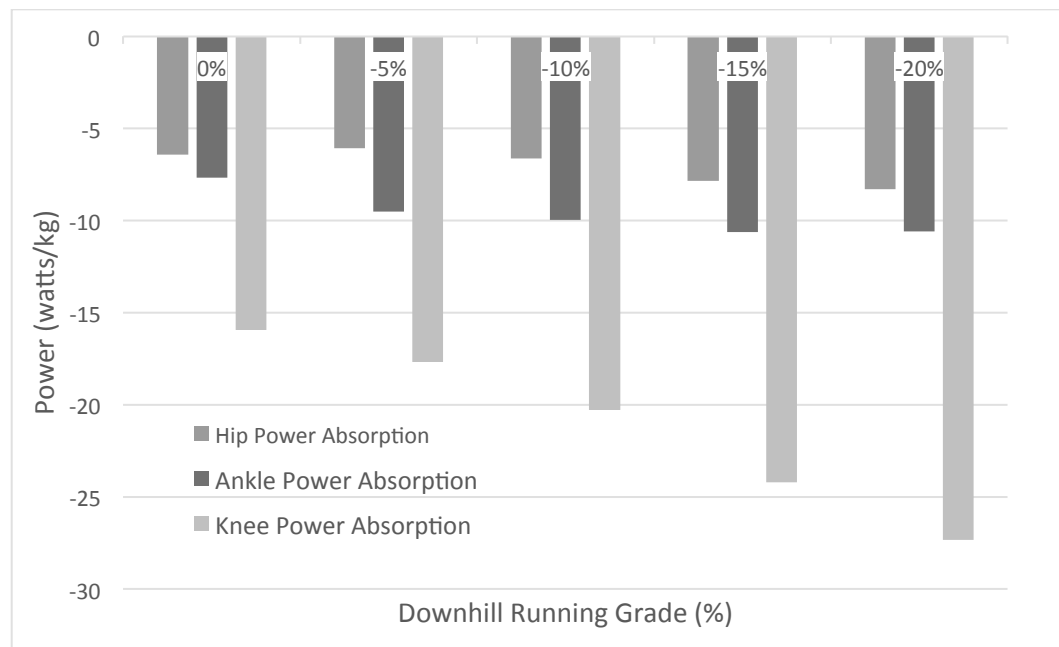


Figure 3: Mean lower extremity power absorption (watts*kg⁻¹) averaged across all subjects for each grade.

Table 2: Mean \pm SD lower extremity joint power (watts*kg⁻¹) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Ankle Absorption	-7.66 \pm 1.21	-9.51* \pm 1.30	-9.98* \pm 2.50	-10.62* \pm 2.23	-10.60* \pm 2.60
Ankle Production	12.41 \pm 2.08	12.73 \pm 1.93	10.97* \pm 1.88	8.97* \pm 1.38	7.06* \pm 1.19
Knee Absorption	-15.95 \pm 2.93	-17.66 \pm 3.41	-20.28* \pm 3.72	-24.20* \pm 4.82	-27.35* \pm 5.48
Knee Production	9.00 \pm 2.08	6.40* \pm 1.68	5.51* \pm 1.45	4.72* \pm 1.14	3.26* \pm 1.11
Hip Absorption	-6.40 \pm 1.68	-6.06 \pm 1.43	-6.61 \pm 1.17	-7.83 \pm 1.76	-8.30* \pm 2.36
Hip Production	2.85 \pm 1.47	2.91 \pm 1.35	2.33 \pm 1.53	2.26 \pm 1.46	2.17 \pm 1.30

(*) Indicates a significant difference from the 0% grade (p<.05)

Table 3: Mean \pm SD sagittal plane joint moments (N*m*kg⁻¹) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Peak Ankle Dorsiflexion	-0.72 \pm 0.26	-0.58 \pm 0.20	-0.63 \pm 0.22	-0.74 \pm 0.19	-0.91 \pm 0.23
Peak Ankle Plantarflexion	2.12 \pm 0.78	2.40* \pm 0.81	2.26* \pm 0.82	2.04 \pm 0.77	1.81* \pm 0.76
Peak Knee Flexion	-0.35 \pm 0.19	-0.43 \pm 0.12	-0.37 \pm 0.16	-0.30 \pm 0.16	-0.27 \pm 0.14
Peak Knee Extension	3.22 \pm 0.35	2.89* \pm 0.30	2.98 \pm 0.33	3.17 \pm 0.38	3.28 \pm 0.43
Peak Hip Flexion	-1.55 \pm 0.37	-1.26* \pm 0.29	-1.27* \pm 0.29	-1.40 \pm 0.37	-1.50 \pm 0.42
Peak Hip Extension	1.75 \pm 0.33	2.11* \pm 0.26	2.09* \pm 0.29	2.11* \pm 0.50	2.16* \pm 0.51

(*) Indicates a significant difference from the 0% grade (p<.05)

Table 4: Mean \pm SD frontal plane joint moments ($\text{N}\cdot\text{m}\cdot\text{kg}^{-1}$) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Peak Ankle Inversion	-0.79 \pm 0.34	-1.06* \pm 0.24	-1.19* \pm 0.26	-1.30* \pm 0.24	-1.29* \pm 0.24
Peak Ankle Eversion	0.06 \pm 0.18	-0.01 \pm 0.25	-0.03 \pm 0.28	0.00 \pm 0.31	0.02 \pm 0.29
Peak Knee Adduction	-0.37 \pm 0.12	-0.60* \pm 0.25	-0.82* \pm 0.23	-1.00* \pm 0.34	-1.14* \pm 0.36
Peak Knee Abduction	0.94 \pm 0.30	0.89 \pm 0.36	0.67* \pm 0.49	0.76* \pm 0.57	0.73* \pm 0.71
Peak Hip Adduction	-0.42 \pm 0.03	-0.53 \pm 0.03	-0.58 \pm 0.04	-0.60* \pm 0.04	-0.70* \pm 0.08
Peak Hip Abduction	2.40 \pm 0.35	2.51 \pm 0.42	2.37 \pm 0.54	2.63 \pm 0.63	2.46 \pm 0.84

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Table 5: Mean \pm SD transverse plane joint moments ($\text{N}\cdot\text{m}\cdot\text{kg}^{-1}$) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Peak Ankle Internal Rotation	-0.36 \pm 0.11	-0.38 \pm 0.08	-0.50* \pm 0.11	-0.64* \pm 0.18	-0.68* \pm 0.23
Peak Ankle External Rotation	0.32 \pm 0.23	0.30 \pm 0.21	0.27 \pm 0.21	0.34 \pm 0.29	0.34 \pm 0.30
Peak Knee Internal Rotation	-0.88 \pm 0.17	-0.88 \pm 0.15	-0.94 \pm 0.19	-1.03* \pm 0.21	-1.01 \pm 0.28
Peak Knee External Rotation	0.27 \pm 0.13	0.36 \pm 0.11	0.38 \pm 0.14	0.44 \pm 0.21	0.47 \pm 0.22
Peak Hip Internal Rotation	-0.37 \pm 0.17	-0.42 \pm 0.18	-0.43 \pm 0.15	-0.49 \pm 0.30	-0.52 \pm 0.33
Peak Hip External Rotation	1.01 \pm 0.18	0.94 \pm 0.30	0.86 \pm 0.29	0.98 \pm 0.42	0.91 \pm 0.51

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Table 6: Mean \pm SD sagittal plane joint angles (degrees) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Ankle Dorsiflexion at Initial Contact	7.84 \pm 5.77	9.07 \pm 3.17	7.66 \pm 6.26	7.99 \pm 4.16	6.61 \pm 4.64
Peak Ankle Dorsiflexion	25.37 \pm 2.22	23.43* \pm 2.53	21.92* \pm 3.18	21.30* \pm 3.00	19.81* \pm 2.86
Peak Ankle Plantarflexion	-14.37 \pm 4.45	-11.30* \pm 4.37	-8.95* \pm 4.58	-4.97* \pm 3.72	-4.85* \pm 3.33
Knee Flexion at Initial Contact	16.72 \pm 5.12	11.71* \pm 4.11	9.76* \pm 4.31	9.26* \pm 4.29	10.13* \pm 4.27
Peak Knee Extension	12.41 \pm 4.53	11.11 \pm 4.30	9.80 \pm 4.39	9.30 \pm 4.28	10.11 \pm 4.36
Peak Knee Flexion	42.38 \pm 4.50	41.52 \pm 4.48	42.36 \pm 5.18	44.39 \pm 5.16	47.80* \pm 6.43
Hip Flexion at Initial Contact	42.42 \pm 7.22	39.12* \pm 5.89	37.48* \pm 5.55	36.15* \pm 5.26	35.45* \pm 6.11
Peak Hip Flexion	42.79 \pm 7.11	39.72* \pm 5.56	38.46* \pm 5.41	38.18* \pm 4.62	38.01* \pm 4.92
Peak Hip Extension	-7.89 \pm 3.79	-7.57* \pm 3.66	-6.38* \pm 3.59	-4.43* \pm 4.26	0.64* \pm 6.66
Trunk Extension at Initial Contact	-14.02 \pm 5.50	-16.94* \pm 5.86	-18.79* \pm 5.61	-19.59* \pm 5.47	-19.53* \pm 5.33

(*) Indicates a significant difference from the 0% grade ($p < .05$)

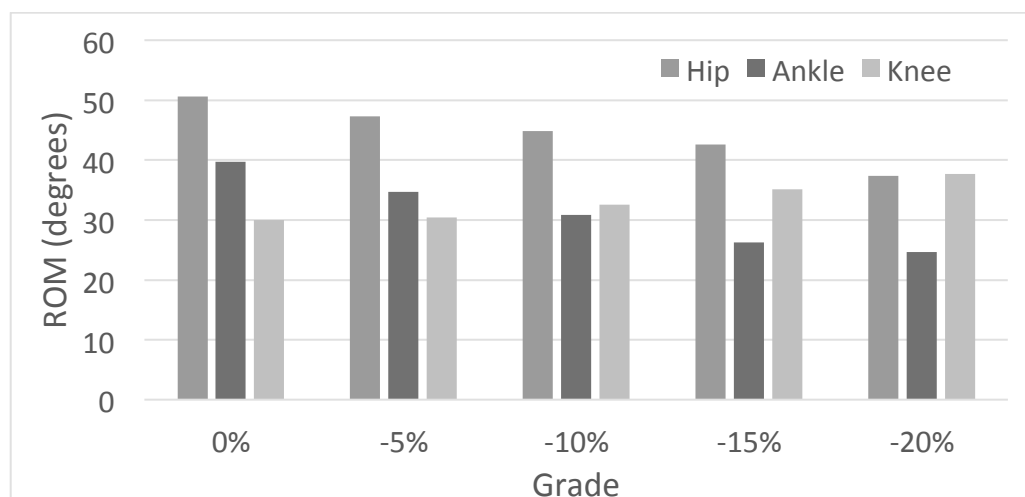


Figure 4: Mean range of motion (degrees) at the ankle, knee, and hip in the stance phase averaged across all subjects for each grade.

Table 7: Mean \pm SD frontal plane joint angles (degrees) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Ankle Inversion at Initial Contact	7.09 \pm 4.65	8.58 \pm 4.61	8.81 \pm 4.33	8.93 \pm 4.43	8.67 \pm 3.95
Peak Ankle Inversion	11.15 \pm 4.38	11.46 \pm 4.10	11.26 \pm 4.04	10.75 \pm 4.21	10.14 \pm 3.69
Peak Ankle Eversion	-11.66 \pm 2.75	-11.25 \pm 2.76	-10.74 \pm 2.93	-10.28* \pm 2.75	-9.52* \pm 2.46
Knee Abduction at Initial Contact	-2.66 \pm 3.30	-2.72 \pm 3.14	-2.67 \pm 3.29	-2.68 \pm 3.33	-2.76 \pm 3.63
Peak Knee Abduction	-4.01 \pm 3.74	-3.95 \pm 3.68	-3.98 \pm 3.81	-3.87 \pm 3.73	-3.58 \pm 3.81
Peak Knee Adduction	0.81 \pm 3.78	1.31 \pm 4.08	1.38 \pm 4.24	1.83 \pm 4.58	2.03 \pm 4.91
Hip Adduction at Initial Contact	11.62 \pm 2.89	9.79* \pm 2.61	7.63* \pm 2.59	5.33* \pm 2.63	3.04* \pm 2.60
Peak Hip Abduction	1.66 \pm 2.12	2.18 \pm 2.04	2.74* \pm 2.16	2.95* \pm 1.96	2.09 \pm 2.02
Peak Hip Adduction	16.88 \pm 2.92	16.23* \pm 2.83	15.18* \pm 2.88	14.18* \pm 2.99	12.20* \pm 2.84

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Table 8: Mean \pm SD transverse plane joint angles (degrees) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Peak Ankle External Rotation	-18.62* \pm 7.38	-19.20* \pm 7.28	-19.57* \pm 7.17	-20.47* \pm 6.82	-20.41* \pm 6.82
Peak Ankle Internal Rotation	-7.47 \pm 9.27	-7.71 \pm 9.37	-8.08 \pm 8.71	-8.81 \pm 8.91	-9.55 \pm 9.00
Peak Knee External Rotation	-6.20 \pm 6.18	-6.78 \pm 6.71	-6.42 \pm 7.14	-6.13 \pm 6.66	-5.21 \pm 7.43
Peak Knee Internal Rotation	7.29 \pm 5.91	6.99 \pm 6.28	7.36 \pm 6.37	7.61 \pm 6.17	7.50 \pm 6.19
Peak Hip External Rotation	-5.20 \pm 7.17	-5.35 \pm 6.27	-6.00 \pm 5.83	-6.81 \pm 5.87	-7.80* \pm 6.21
Peak Hip Internal Rotation	3.82 \pm 7.13	3.91 \pm 7.17	3.06 \pm 7.39	2.78* \pm 7.22	1.07* \pm 7.09

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Table 9: Mean \pm SD spatiotemporal parameters averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Stride Rate (per minute)	92.00 \pm 4.05	91.03 \pm 4.50	90.55 \pm 5.20	91.04 \pm 5.75	94.48 \pm 6.60
Step Length (m)	1.31 \pm 0.05	1.32 \pm 0.06	1.33 \pm 0.07	1.32 \pm 0.07	1.28 \pm 0.08

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Table 10: Mean \pm SD average and peak muscle activity during stance as a percentage of the MVC averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Gluteus Medius (Avg)	38.79 \pm 38.07	39.80 \pm 38.16	73.62 \pm 167.67	57.38 \pm 88.32	60.42 \pm 73.49
Gluteus Medius (Peak)	121.13 \pm 150.34	97.42 \pm 145.34	157.26 \pm 372.88	116.93 \pm 182.78	121.13 \pm 150.34
Biceps Femoris (Avg)	35.55 \pm 18.73	39.76 \pm 34.67	26.85 \pm 13.64	31.64 \pm 23.72	30.55 \pm 18.38
Biceps Femoris (Peak)	59.64 \pm 38.81	72.78 \pm 45.93	52.59 \pm 23.41	57.94 \pm 33.11	59.64 \pm 38.81
Vastus Medialis(Avg)	79.53 \pm 34.28	83.62 \pm 53.86	98.53 \pm 79.22	105.79 \pm 79.40	101.48 \pm 78.29
Vastus Medialis (Peak)	185.44 \pm 138.43	159.23 \pm 95.29	184.75 \pm 154.83	206.31 \pm 185.87	185.44 \pm 138.43
Tibialis Anterior (Avg)	33.65 \pm 23.07	35.62 \pm 28.60	45.90 \pm 47.99	37.12 \pm 17.58	47.58 \pm 34.57
Tibialis Anterior (Peak)	91.92 \pm 91.92	74.35 \pm 51.74	97.16 \pm 93.82	72.38 \pm 26.61	91.91 \pm 62.56
Medial Gastrocnemius (Avg)	86.07 \pm 38.51	79.42 \pm 38.13	69.25* \pm 35.38	59.22* \pm 24.27	55.08* \pm 24.31
Medial Gastrocnemius (Peak)	95.24 \pm 40.36	128.84* \pm 47.93	120.25 \pm 57.05	103.45 \pm 41.31	95.24 \pm 40.36

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Chapter 5

Research Article

The following paper was written with the expectation that it would be submitted to the Journal of Applied Biomechanics for review following completion of the thesis process.

May, 2017

JAB_2017_

**The Effect of Grade on Joint Mechanics and Muscle Activity During Downhill Running in
Female Distance Runners**

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Conflict of Interest Disclosure: None

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Abstract

Hill running, both up and down, is often used as a foundational training mechanism to build strength and speed. There is limited research available regarding downhill running, and there is no research available, that I know of, on the biomechanics of steep hill running. The purpose of this study was to quantify the biomechanics of downhill running at four different grades compared to level in female distance runners, and to determine the potential injury risk when running downhill. Fifteen healthy, female distance runners, ages 18-35, who ran a minimum of 15 miles per week participated in this study. Participants ran on a force-instrumented treadmill at 4.0 m/s for 2 minutes on 0%, -5%, -10%, -15%, and -20% grades in a random order, with 5 minutes of rest between each. Key study findings were increased power absorption at the ankle and knee ($p<.002$), decreased hip and knee flexion and increased trunk extension at initial contact ($p<.001$), and increased knee ($p<.001$) and hip ($p=.036$) adduction moments. Based on risk factors for common running injuries illustrated in prior research, the results from this study indicate that there is a greater risk for injury when running downhill compared to running on a level surface.

Keywords: distance running, lower extremity, biomechanics, injury risk, overuse injury

Word Count: 3,819

Introduction

Running is a popular activity among individuals of all ages, ethnicities, and economic backgrounds. Physical fitness activities, such as running, can aid in the prevention of stroke, heart disease, diabetes, and osteoporosis ¹. Running continues to grow in popularity as the medical focus moves from treating disease to preventing disease ^{1,2}. Long distance running in particular is practiced by many people because it requires little to no skill, is low cost, and can be done anywhere. Long distance runners have an increased likelihood of running up and down hills while out on long runs, and competitive runners use hill running, both up and down, as a key training technique to build strength and speed ³. Additionally, many distance runners train on trails, which often introduces more inclines and declines than running on paved surfaces.

Although the benefits of running are numerous, there is also significant chance for injury at some point in a runner's career, especially to the lower extremity ^{2,4}. Running is one of the most common activities in which musculoskeletal overuse injury of the lower extremity occurs ⁵. Various studies have found that over the course of a year anywhere from 27 to 79 percent of distance runners will experience an overuse injury ^{4,6,7}, the majority of which occur at the knee ⁶⁻⁸. Musculoskeletal overuse injuries arise as a result of combined fatigue over an extended period of time beyond the capabilities of a specific structure such as muscle, tendon, or bone ⁹. Additionally, females are at a greater risk for injury than males due to anatomical and biomechanical differences ^{8,10}.

Previous research on downhill running has shown that running downhill results in less hip and knee flexion at initial contact, greater peak knee flexion, and a greater ROM at the knee ¹¹⁻¹³. The motion at the knee that has been seen in these previous studies has been associated with common knee injuries such as PFPS and ITBFS ^{8,14,15}.

Prior research regarding power absorption, joint moments, and muscle activity has been very conflicted. One study reported that the power absorption at the knee and ankle increased significantly with downhill running ¹¹, while a second study reported no differences in power absorption at the ankle, but greater power absorption at the hip ¹⁶. Additionally, the peak plantar flexor moment and peak hip extensor moment had been reported to be smaller when running downhill ¹², yet another study stated no differences in joint moments ¹⁶.

In addition, two previous studies analyzing downhill running reported no differences in spatiotemporal parameters ^{12,17}, while a third discovered that the stride rate decreased from level running to downhill running ¹⁸. The muscle groups that seem to be most affected by downhill running are the gluteal muscles, the quadriceps, and the calf muscles, all of which are the extensor muscles of the hip, knee, and ankle ¹¹. However, it has yet to be established how the muscle activity in the lower extremity changes when the grade is declined.

Most running injuries likely occur due to downhill running, however, the mechanics of downhill running are not well understood, especially when it comes to steeper declines. Although numerous studies have been conducted on level running, there is limited research available regarding downhill running. Additionally, previous research has been concentrated at low grades, and it is unknown what happens biomechanically when the decline becomes greater. More research is needed in order to understand the biomechanics associated with steep downhill running grades because this running condition occurs frequently during distance training. Therefore, the purpose of this study was to quantify the differences in mechanics and muscle activity for downhill running at the four different grades compared to level running in female distance runners, and to determine if there was a dose-response relationship between the grade and the biomechanical variables, as well as to analyze female distance runners' lower extremity

joint mechanics, spatiotemporal parameters, and muscle activity in order to determine the potential risk for knee injury. It was hypothesized that as the decline became greater there would be decreased hip and knee flexion at ground contact, greater moments at the ankle and knee, altered muscle activity, and a decreased stride rate with an increased stride length.

Methods

Fifteen female distance runners (age: 23.5 ± 4.9 y, Height: $1.7 \pm .06$ m, Weight: 57.8 ± 6.8 kg) participated in this study. Participants were regular runners who trained an average of 35 ± 13 miles a week. Participants were in good cardiovascular health, had no history of lower extremity injury in the six months prior to the study, and were free of any chronic musculoskeletal conditions or neurological problems that would inhibit their ability to run normally. Prior to data collection, IRB approval was obtained and each participant read and signed an informed consent.

Participants filled out health and running history questionnaires to determine eligibility, and were provided with compression clothing and standardized Nike shoes (NIKE, Inc., Portland, OR, USA). Anthropometric measurements of the lower extremity were taken, wireless EMG sensors (Delsys Inc., Boston, MA, USA) were placed on the belly of the gluteus medius, vastus medialis, biceps femoris, medial gastrocnemius, and tibialis anterior on the dominant limb, and retroreflective markers were placed on anatomical landmarks following a modified Plug-in-Gait marker set.

Participants warmed up on a level grade for five minutes at a comfortable pace and then were fit with a harness attached to the ceiling as a safety precaution. Participants ran for two minutes at 4.0 m/s on five different grades (0%, -5%, -10%, -15%, and -20%) in a randomized order with five minutes of rest between each running condition. Participants then completed

maximum voluntary isometric contractions at the hip, knee, and ankle on a Cybex Dynamometer (Cybex International, NY, USA) sampling at 100 Hz for the purpose of normalizing the EMG during running. The maximum voluntary isometric contractions performed were hip abduction, knee flexion and extension, and ankle plantarflexion and dorsiflexion. Trials lasted for three seconds with a minute of rest between each.

All running took place on an AMTI force instrumented treadmill, sampling at 2000 Hz, controlled by an End-to-End software (Advanced Mechanical Technology, Inc., Watertown, MA, USA). Motion capture data were collected using 15 VICON infrared cameras (Oxford Metrics, Oxford, UK) sampling at 200 Hz. Visual 3D v.5 (C-Motion, Inc., Germantown, MD, USA) was used to calculate peak joint angles and moments in three planes, joint powers, spatiotemporal parameters, and muscle activity averaged over 5 foot strikes from 3 trials for each grade. A customized Matlab program was used to calibrate the treadmill gradients. Statistical significance was set at $\alpha=0.05$, and was determined using repeated measures MANOVAs in SPSS v.24 (SPSS Inc., Chicago, IL, USA). If the main MANOVA showed significance, ANOVA results were then examined for significance. If the univariate tests showed significance, pairwise comparisons with a Bonferroni correction factor were further utilized to determine which conditions the significant differences occurred between. The primary variables were power absorption, and joint angles and moments in all three planes at all three lower extremity joints. Secondary variables were joint power propulsion, lower extremity muscle activity, and spatiotemporal parameters.

Results

Each of the repeated measure MANOVAs showed significant results ($p<.05$). There were significant differences in ankle ($p=.002$), knee ($p<.001$), and hip ($p<.001$) power absorption

(Figure 1) and in the ankle ($p<.001$) and knee ($p<.001$) power propulsion. Ankle power absorption increased in each of the downhill grades compared to level. Knee power absorption was also greater in each of the downhill grades compared to level, however, 0% and -5% were not significantly different from each other. There was also a dose-response in 10% increments, so -10% was greater than 0% ($p=.001$), -15% was greater than -5% ($p<.001$), and -20% was greater than -10% ($p<.001$). Hip power absorption increased from 0% to -20% ($p=.044$). Power propulsion decreased in a dose-response relationship at both the ankle and knee.

There were significant differences in peak flexion and extension moments at the ankle, knee, and hip among the downhill grades ($p<.01$). However, the differences in the downhill grades compared to level were minimal (Table 1). The peak ankle inversion moment was greater in each of the downhill grades compared to level ($p<.001$). The peak knee adduction moment was greater in each of the downhill grades compared to level, and there was a dose response relationship, but -10% and -15% were not significantly different. The -5% grade was greater than 0% ($p=.004$), -10% was greater than -5% ($p=.004$), and -20% was greater than -10% ($p=.019$) and -15% ($p=.005$). The peak knee abduction moment decreased from 0% to -10% ($p=.001$), 0% to -15% ($p=.010$), and 0% to -20% ($p=.005$). The peak hip adduction moment increased from 0% to -15% ($p=.036$), and 0% to -20% ($p=.042$). There were no differences seen in the peak ankle eversion, peak knee abduction, or peak hip abduction moments ($p>.05$) (Table 2). The peak ankle internal rotation moment increased from 0% to -10% ($p=.003$), 0% to -15% ($p<.001$), and 0% to -20% ($p=.001$). Peak knee internal rotation was significantly greater in the -15% condition compared to 0% ($p=.011$) and -5% ($p=.002$). There were no differences in the peak knee or ankle external rotation moments, or hip internal or external rotation moments ($p>.05$) (Table 3).

The peak ankle dorsiflexion angle decreased in each of the downhill grades compared to level, and there was a dose-response relationship, but -10% and -15% were not significantly different from each other. The -5% grade was smaller than 0% ($p<.001$), -10% was smaller -5% ($p=.001$), and -20% was smaller than -15% ($p<.001$) and -10% ($p<.001$). The peak plantarflexion angle decreased in each of the downhill grades, and there was a dose-response pattern, however, -15% and -20% were not significantly different from each other. The -5% grade was smaller than 0% ($p=.008$), -10% was smaller than -5% ($p=.001$), and -15% was smaller than -10% ($p=.001$). The knee flexion angle at initial contact decreased (increased extension) between level and each of the downhill grades ($p<.001$). The peak knee flexion angle increased from 0% to -20% ($p=.013$). The hip flexion angle at initial contact decreased from level to each of the downhill grades ($p<.001$). The peak hip extension angle decreased in each of the downhill grades (except -5%) compared to level, and there was a dose-response relationship. The -10% grade had a smaller peak hip extension angle than both 0% ($p=.033$) and -5% ($p=.026$), -15% was smaller than -10% ($p=.010$), and -20% was smaller than -15% ($p=.014$). The trunk extension at initial contact increased from level to each of the downhill grades ($p<.001$). The changes in lower extremity joint angles from level to downhill resulted in a decreased ROM at the ankle and hip, and an increased ROM at the knee in the sagittal plane (Figure 2).

The peak ankle eversion angle decreased from 0% to -15% ($p=.001$), and 0% to -20% ($p<.001$). The hip adduction angle at initial contact decreased in each of the downhill grades compared to level, and there was a dose-response relationship. The -5% grade was smaller than 0% ($p<.001$), -10% was smaller than -5% ($p<.001$), -15% was smaller than -10% ($p<.001$), and -20% was smaller than -15% ($p<.001$). The peak hip adduction angle also decreased in each of the downhill grades, and there was a dose-response relationship seen, however, -10% and -15% were

not significantly different from each other. The angle decreased from 0% to -5% ($p=.034$), -5% to -10% ($p=.034$), -10% to -20% ($p<.001$), and from -15% to -20% ($p<.001$). The hip never became abducted, but the minimum adduction angle increased from 0% to -10% ($p=.007$) and from 0% to -15% ($p=.007$) (Table 4).

The peak ankle external rotation angle increased from 0% to -5% ($p=.030$), 0% to -10% ($p=.024$), 0% to -15% ($p=.003$), and from 0% to -20% ($p=.002$). The peak hip external rotation angle increased from 0% to -20% ($p=.002$). from -5% to -20% ($p<.001$), from -10% to -20% ($p=.001$), and from -15% to -20% ($p=.003$). The peak hip internal rotation angle decreased from 0% to -15% ($p=.025$), and from 0% to -20% ($p<.001$) (Table 5).

There were no significant differences in stride rate or stride length at any of the grades ($p>.05$). The average medial gastrocnemius muscle activity decreased from 0% to -10% ($p=.012$), 0% to -15% ($p=.017$), and 0% to -20% ($p=.024$) (Table 6).

Discussion

The purpose of this study was to quantify the differences in biomechanics for downhill running at five different grades in female distance runners, and to analyze female distance runners' lower extremity joint mechanics, spatiotemporal parameters, and muscle activity in order to determine the potential risk for lower extremity injury. The results from this study partially supported the hypotheses. The main findings of this study included increased power absorption at the knee and ankle, decreased knee and hip flexion and increased trunk extension at initial contact, and increased knee and hip joint adduction moments.

Previous research has reported conflicting research regarding power absorption in the lower extremity^{11,12,16}. In the present study, each of the downhill grades illustrated significantly greater ankle power absorption compared to level, and the amount of power absorbed increased

by almost half between level and -20%. The knee power absorption nearly doubled between the level condition and the -20% condition. The increased knee power absorption is consistent with previous research ¹¹, however, the present study also established that there is a dose-response relationship between the decline and the amount of power absorbed by the knee joint. This is thought to be the reason for the significant soreness felt by runners when training downhill ^{12,16}. A significant increase in power absorbed by the hip was also observed in the present study, although to a lesser extent than at the knee and ankle. Among the three joints, the knee absorbed considerably more power at each of the grades. Additionally, the difference in power absorption between the joints became increasingly greater as the decline became greater. This may be an indication that the knee is at a greater risk for injury when running downhill compared to the ankle and hip.

There was an increase in the peak plantar flexor moment in the -5% and -10% conditions compared to level, however, at -15% and -20% the peak plantar flexor moment was smaller than in the level condition, which was not expected. However, the decreased plantar flexor moment does correlate to the decrease seen in the medial gastrocnemius muscle activity. The knee extension moment during the level condition was greater than in the -5% condition, but there were no other significant differences between the downhill grades when compared to the level condition. There also were no differences seen in the knee flexion moment. Although this was a surprising result, it does correlate to the absence of changes seen in the biceps femoris and vastus medialis muscle activity. This is similar to what has been found in the past ¹¹, however, it was unexpected because the muscle soreness that is often reported by runners is generally in the quadriceps muscles, or the knee extensors ¹¹.

The knee adduction moment increased significantly in each of the downhill grades compared to level, and there was also a dose-response relationship that illustrated an increased adduction moment as the grade increased. This is an indication that there is increased lateral stress being placed on the knee. The increased stress on the lateral aspect of the knee may lead to an increased risk for lateral knee pain, which is a common complaint with ITBFS⁸. There was an increased adduction moment at the hip in the -15% and -20% grades as well, which correlates to the increased adduction moment seen at the knee. The increased hip adduction moment may lead to overuse of the hip adductor muscles, which in turn may cause fatigue and muscle soreness in the hip adductor muscles. However, the increased hip adduction moment could result in the hip adductor muscles becoming stronger.

Our results showed less knee and hip flexion at initial contact during downhill running, which may lead to greater impact forces. There was also a greater peak knee flexion angle. The results observed at the knee are identical to what was reported by Buczek and Cavanagh (1990) and Mizrahi, Verbitsky, and Isakov (2000). However, both of these previous studies used only male runners, and used much smaller declines. The decreased knee flexion at initial contact is likely the result of runners having to extend further in order to reach for the lower ground¹², and the resulting increased ROM at the knee has been attributed to needing to cushion the landing more¹¹. Additionally, the present study also established a dose-response relationship between the downhill grade and the resulting ROM at each of the lower extremity joints. As the downhill grade became steeper, the ROM at both the ankle and the hip decreased, and the ROM at the knee increased. Additionally, there was a greater amount of trunk extension when running downhill, similar to previous research¹⁸, indicating that runners lean backward more when running downhill. The decreased ROM at the ankle may be linked to the decreased power

production at the ankle, the decreased plantarflexion moment, and the decreased muscle activity in the medial gastrocnemius muscle. The increased ROM at the knee when running downhill likely contributes to ITBFS and PFPS in distance runners^{14,15}. Runners have reported that running downhill increases irritation of the condition¹⁹, which could be the result of downhill running involving landing in a more extended position and moving through a greater ROM in the stance phase¹⁴. Anterior knee pain, as occurs with PFPS, is also heightened by increased flexion and extension at the knee¹⁵, so downhill running may also increase the risk of PFPS in runners.

There were decreased ankle eversion angles seen in the downhill grades. Ankle eversion is one of the three combination movements involved in pronation. Pronation allows for forces to be attenuated over a longer period of time, and thus serves as a protective mechanism during running^{9,20}. As the running grade increased, the eversion angle decreased, and the ankle became stiffer. This is an indication that the forces were not being attenuated as well, which could increase the risk for injuries such as stress fractures. Additionally, results from the present study illustrated increased hip external rotation in the steeper declines, and the peak internal rotation also decreased, so the hip was in a greater amount of external rotation throughout stance. Greater external rotation at the hip will place increased stress on the distal end of the IT band, which may increase the risk for ITBFS. Previous research has indicated that females who develop ITBFS, or who have a history of ITBFS, exhibit greater femoral external rotation when running^{21,22}. Greater hip external rotation, without a change in knee external rotation means the runners were landing more on the lateral aspect of the foot, which correlates with the decrease in eversion seen in this study as well.

Changes in speed result from changes in either the stride rate or the step length. In order to run faster either the stride length and/or the step length must increase, and in order to run

slower one or both must decrease. It was hypothesized that as the downhill grade became steeper that the stride rate would decrease and step length would increase. Previous research has reported conflicting results regarding these spatiotemporal parameters. One study found a decreased stride rate¹⁸, while others found an increased stride length during downhill running²³, and still others found no differences at all in the stride rate or the step length^{12,17}. The present study showed no changes in either the stride rate or the step length. Although differences in stride rate and step length were expected, the lack of differences found may have resulted from the constant running speed among all of the conditions.

Previous research indicated that the most active muscles during downhill running are the gluteal muscles, the quadriceps, and the triceps surae, which are the extensor muscles of the lower extremity¹¹. Downhill running has been found to be connected to the greatest onset of delayed onset muscle soreness compared to level and uphill running, likely due to an increase in eccentric muscle activity that is required when running downhill (Brown et al., 1999; Cai et al., 2010). It was hypothesized that there would be changes in the level of muscle activation when running downhill, however, it was unpredictable as to what those changes would be because of conflicting results in past research^{11,25}. However, the only difference observed was decreased muscle activity in the medial gastrocnemius muscle. Although surprising, it is consistent with other findings. The decreased ankle ROM and plantarflexion moment, could in part be related to the reduced gastrocnemius activity at the steeper grades. Additionally, the more rigid ankle joint could be linked to the decrease in power production at the ankle, which would also explain the decrease in muscle activity seen in the gastrocnemius muscle.

There are a few limitations to this study. Participants all ran in standardized running shoes in order to eliminate differences that could occur due to worn out shoes or any orthotic

inserts or specially made shoes. This could have potentially introduced differences in mechanics due to being unfamiliar with the type of shoes used. However, everyone wore the same shoes, so any changes that occurred should have been relatively consistent across all participants. Additionally, this study was conducted in a controlled laboratory setting, using a set running speed.

Previous research on downhill running has been limited, and there are many conflicting results in the studies that have been done. The present study is, to the best of our knowledge, the first study to examine downhill running on grades similar to those that would be encountered on hilly trail runs. It is also believed to be the first study that has examined whether a dose-response relationship exists between decline and joint mechanics, and the first study to analyze female mechanics in downhill running.

The most significant findings of this study include increased power absorption at the knee and ankle, decreased hip and knee flexion and increased trunk extension at initial contact, and increased knee and hip joint adduction moments. The increased ROM at the knee and the decreased knee flexion at initial contact are risk factors for knee pain such as in PFPS and ITBFS. The increased knee and hip adduction moments may lead to increased lateral knee pain. Overall, there appears to be a greater risk for sustaining an injury when running downhill. However, this was not a prospective study so it cannot be stated that downhill running causes injuries. Additionally, downhill running is a necessary part of distance training, and should not be completely eliminated. Rather, runners and coaches need to be aware of the potential risks associated with downhill running, so that they may be taken into account when creating training programs so that injuries may be avoided.

Tables and Figures

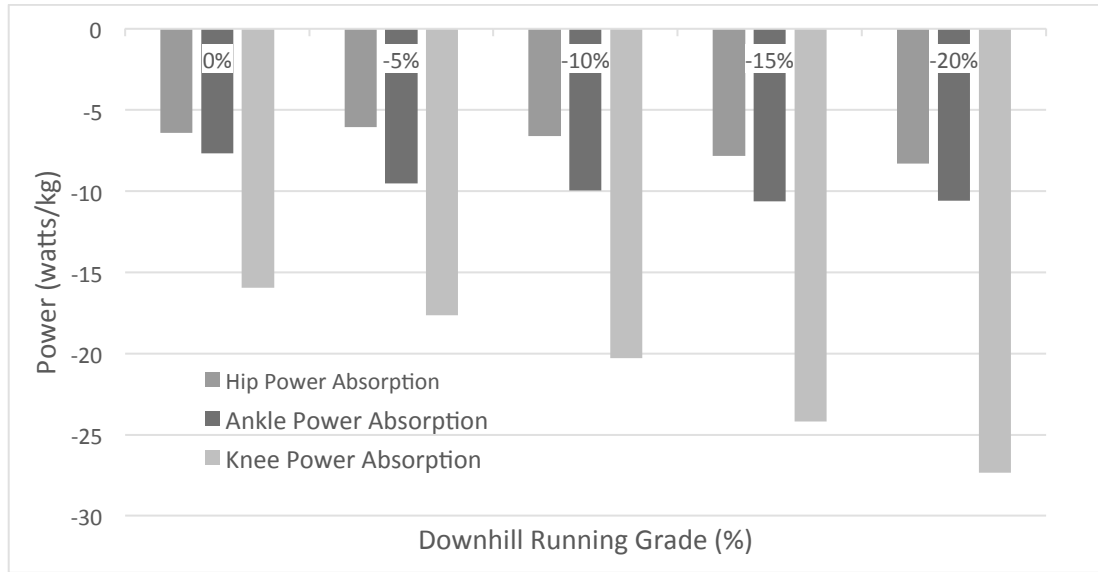


Figure 1: Mean lower extremity power absorption ($\text{watts} \cdot \text{kg}^{-1}$) averaged across all subjects for each grade.

Table 1: Mean \pm SD sagittal plane joint moments ($\text{N} \cdot \text{m} \cdot \text{kg}^{-1}$) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Peak Ankle Dorsiflexion	-0.72 \pm 0.26	-0.58 \pm 0.20	-0.63 \pm 0.22	-0.74 \pm 0.19	-0.91 \pm 0.23
Peak Ankle Plantarflexion	2.12 \pm 0.78	2.40* \pm 0.81	2.26* \pm 0.82	2.04 \pm 0.77	1.81* \pm 0.76
Peak Knee Flexion	-0.35 \pm 0.19	-0.43 \pm 0.12	-0.37 \pm 0.16	-0.30 \pm 0.16	-0.27 \pm 0.14
Peak Knee Extension	3.22 \pm 0.35	2.89* \pm 0.30	2.98 \pm 0.33	3.17 \pm 0.38	3.28 \pm 0.43
Peak Hip Flexion	-1.55 \pm 0.37	-1.26* \pm 0.29	-1.27* \pm 0.29	-1.40 \pm 0.37	-1.50 \pm 0.42
Peak Hip Extension	1.75 \pm 0.33	2.11* \pm 0.26	2.09* \pm 0.29	2.11* \pm 0.50	2.16* \pm 0.51

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Table 2: Mean \pm SD frontal plane joint moments ($\text{N}\cdot\text{m}\cdot\text{kg}^{-1}$) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Peak Ankle Inversion	-0.79 \pm 0.34	-1.06* \pm 0.24	-1.19* \pm 0.26	-1.30* \pm 0.24	-1.29* \pm 0.24
Peak Ankle Eversion	0.06 \pm 0.18	-0.01 \pm 0.25	-0.03 \pm 0.28	0.00 \pm 0.31	0.02 \pm 0.29
Peak Knee Adduction	-0.37 \pm 0.12	-0.60* \pm 0.25	-0.82* \pm 0.23	-1.00* \pm 0.34	-1.14* \pm 0.36
Peak Knee Abduction	0.94 \pm 0.30	0.89 \pm 0.36	0.67* \pm 0.49	0.76* \pm 0.57	0.73* \pm 0.71
Peak Hip Adduction	-0.42 \pm 0.03	-0.53 \pm 0.03	-0.58 \pm 0.04	-0.60* \pm 0.04	-0.70* \pm 0.08
Peak Hip Abduction	2.40 \pm 0.35	2.51 \pm 0.42	2.37 \pm 0.54	2.63 \pm 0.63	2.46 \pm 0.84

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Table 3: Mean \pm SD transverse plane joint moments ($\text{N}\cdot\text{m}\cdot\text{kg}^{-1}$) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Peak Ankle Internal Rotation	-0.36 \pm 0.11	-0.38 \pm 0.08	-0.50* \pm 0.11	-0.64* \pm 0.18	-0.68* \pm 0.23
Peak Ankle External Rotation	0.32 \pm 0.23	0.30 \pm 0.21	0.27 \pm 0.21	0.34 \pm 0.29	0.34 \pm 0.30
Peak Knee Internal Rotation	-0.88 \pm 0.17	-0.88 \pm 0.15	-0.94 \pm 0.19	-1.03* \pm 0.21	-1.01 \pm 0.28
Peak Knee External Rotation	0.27 \pm 0.13	0.36 \pm 0.11	0.38 \pm 0.14	0.44 \pm 0.21	0.47 \pm 0.22
Peak Hip Internal Rotation	-0.37 \pm 0.17	-0.42 \pm 0.18	-0.43 \pm 0.15	-0.49 \pm 0.30	-0.52 \pm 0.33
Peak Hip External Rotation	1.01 \pm 0.18	0.94 \pm 0.30	0.86 \pm 0.29	0.98 \pm 0.42	0.91 \pm 0.51

(*) Indicates a significant difference from the 0% grade ($p < .05$)

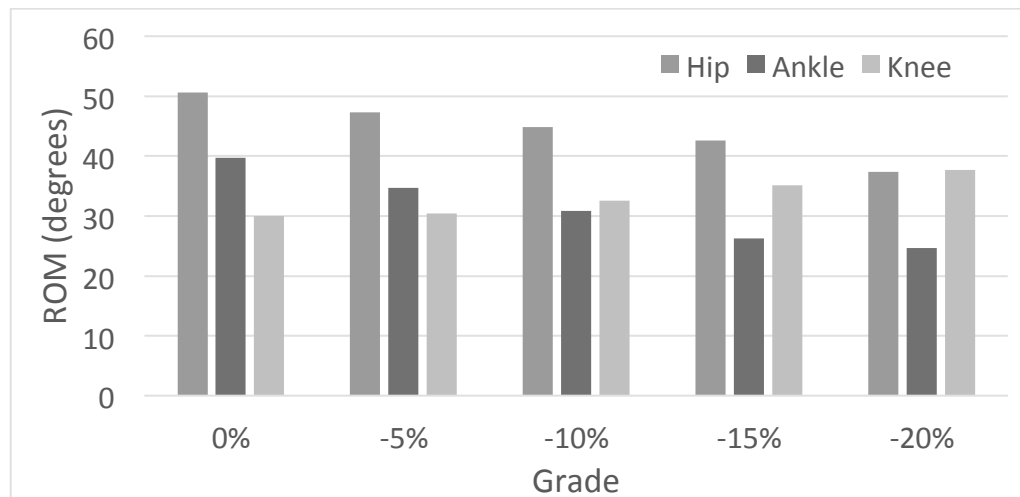


Figure 2: Mean range of motion (degrees) at the ankle, knee, and hip in the stance phase averaged across all subjects for each grade.

Table 4: Mean \pm SD frontal plane joint angles (degrees) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Ankle Inversion at Initial Contact	7.09 \pm 4.65	8.58 \pm 4.61	8.81 \pm 4.33	8.93 \pm 4.43	8.67 \pm 3.95
Peak Ankle Inversion	11.15 \pm 4.38	11.46 \pm 4.10	11.26 \pm 4.04	10.75 \pm 4.21	10.14 \pm 3.69
Peak Ankle Eversion	-11.66 \pm 2.75	-11.25 \pm 2.76	-10.74 \pm 2.93	-10.28* \pm 2.75	-9.52* \pm 2.46
Knee Abduction at Initial Contact	-2.66 \pm 3.30	-2.72 \pm 3.14	-2.67 \pm 3.29	-2.68 \pm 3.33	-2.76 \pm 3.63
Peak Knee Abduction	-4.01 \pm 3.74	-3.95 \pm 3.68	-3.98 \pm 3.81	-3.87 \pm 3.73	-3.58 \pm 3.81
Peak Knee Adduction	0.81 \pm 3.78	1.31 \pm 4.08	1.38 \pm 4.24	1.83 \pm 4.58	2.03 \pm 4.91
Hip Adduction at Initial Contact	11.62 \pm 2.89	9.79* \pm 2.61	7.63* \pm 2.59	5.33* \pm 2.63	3.04* \pm 2.60
Peak Hip Abduction	1.66 \pm 2.12	2.18 \pm 2.04	2.74* \pm 2.16	2.95* \pm 1.96	2.09 \pm 2.02
Peak Hip Adduction	16.88 \pm 2.92	16.23* \pm 2.83	15.18* \pm 2.88	14.18* \pm 2.99	12.20* \pm 2.84

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Table 5: Mean \pm SD transverse plane joint angles (degrees) averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Peak Ankle External Rotation	-18.62* \pm 7.38	-19.20* \pm 7.28	-19.57* \pm 7.17	-20.47* \pm 6.82	-20.41* \pm 6.82
Peak Ankle Internal Rotation	-7.47 \pm 9.27	-7.71 \pm 9.37	-8.08 \pm 8.71	-8.81 \pm 8.91	-9.55 \pm 9.00
Peak Knee External Rotation	-6.20 \pm 6.18	-6.78 \pm 6.71	-6.42 \pm 7.14	-6.13 \pm 6.66	-5.21 \pm 7.43
Peak Knee Internal Rotation	7.29 \pm 5.91	6.99 \pm 6.28	7.36 \pm 6.37	7.61 \pm 6.17	7.50 \pm 6.19
Peak Hip External Rotation	-5.20 \pm 7.17	-5.35 \pm 6.27	-6.00 \pm 5.83	-6.81 \pm 5.87	-7.80* \pm 6.21
Peak Hip Internal Rotation	3.82 \pm 7.13	3.91 \pm 7.17	3.06 \pm 7.39	2.78* \pm 7.22	1.07* \pm 7.09

(*) Indicates a significant difference from the 0% grade ($p < .05$)

Table 6: Mean \pm SD average muscle activity during stance as a percentage of the MVC averaged across all subjects for each grade.

	0%	-5%	-10%	-15%	-20%
Gluteus Medius	38.79 \pm 38.07	39.80 \pm 38.16	73.62 \pm 167.67	57.38 \pm 88.32	60.42 \pm 73.49
Biceps Femoris	35.55 \pm 18.73	39.76 \pm 34.67	26.85 \pm 13.64	31.64 \pm 23.72	30.55 \pm 18.38
Vastus Medialis	79.53 \pm 34.28	83.62 \pm 53.86	98.53 \pm 79.22	105.79 \pm 79.40	101.48 \pm 78.29
Tibialis Anterior	33.65 \pm 23.07	35.62 \pm 28.60	45.90 \pm 47.99	37.12 \pm 17.58	47.58 \pm 34.57
Medial Gastrocnemius	86.07 \pm 38.51	79.42 \pm 38.13	69.25* \pm 35.38	59.22* \pm 24.27	55.08* \pm 24.31

(*) Indicates a significant difference from the 0% grade ($p < .05$)

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Discussion and Conclusions

Discussion

The primary purpose of this study was to quantify the differences in mechanics and muscle activity for downhill running at the four different grades compared to level running in female distance runners, and to determine if there was a dose-response relationship between the mechanics and the downhill grade. The secondary purpose of this study was to analyze female distance runners' lower extremity joint kinematics, kinetics, spatiotemporal parameters, and muscle activity in order to determine if there is a greater potential for injury when running down different declines.

In general, it was hypothesized that there would be greater vertical and braking GRFs, and decreased propulsive forces, as well as increased lower extremity joints moments, and greater hip and knee joint extension at initial contact as the grades became steeper. It was also hypothesized that there would be an increase in stride rate and a decrease in step length, in addition to changes in the muscle activity of the lower extremity.

The secondary hypothesis was that the variables described above would change between the level condition and each of the downhill grades, and that there would also be a dose-response

relationship seen. It was expected that as the downhill grades became steeper, that the differences in the variables would become more exaggerated when compared to a level grade.

These hypotheses were partially supported, as there were increased vertical GRFs and loading rates as the downhill grade increased. There were decreased propulsive forces seen, but there were no changes in the braking force. There were significantly greater forces seen at each of the lower extremity joints, and there was greater extension observed at the hip and knee at initial contact, as expected. Lastly, there were no significant changes in the stride rate or the step length or in the muscle activity, contrary to what was expected.

Ground Reaction Forces

Running is a highly repetitive activity that involves very large forces being applied to the body at a rate of approximately 90 times per minute (Lorimer & Hume, 2014). During heel-toe running, the peak force is generally between two and three times the individual's body weight (Cavanagh & LaFortune, 1980), which can lead to potential bone deformation. The higher the GRF, the GRF loading rate, or the number of repetitions, the greater the bone deformation will be (Bennell et al., 2004). Changing foot contact patterns generally causes the GRF to increase during downhill running (Shorten & Winslow, 1992). In the present study, there were eleven runners that ran with a rear foot strike, and four runners that ran with either a forefoot strike or a mid-foot strike naturally. However, once the grade declined, all of the participants adapted a rear foot strike pattern. It was expected that adapting a rear foot strike would cause the vertical peak GRF or the peak braking force to increase, however, this was not the case. The vertical GRF did not increase until the -10% grade, and the braking forces did not change at all. The level grade showed a peak vertical GRF of 2.5 BWs, and as the grade became steeper the GRF increased until reaching 3.0 BWs at -20%. However, during level running the peak GRF occurred during

the active peak, and in the steeper downhill grades the peak GRF occurred during the impact peak, and the second peak disappeared. Although these are fairly average GRFs in running, the increased impact peak going downhill may have an effect on the risk for potential injury.

Downhill running has been found to produce significantly different GRFs compared to level and uphill running. Since GRFs are a likely contributor to numerous running injuries, including PFPS and stress fractures, it is important to be aware of the impact that the changing grade has on the resulting GRFs. The finding that the vertical peak GRF increases when running down steeper grades is consistent with previous research (Gottschall & Kram, 2005; Telhan et al., 2010; Yokozawa et al., 2005). However, our study is unique from previous research looking at GRFs because we analyzed downhill conditions that are closer to those that would be encountered by distance runners. All three of the prior studies looking at vertical GRFs analyzed grades that are the equivalent of rolling hills, while the present study analyzed grades that are much steeper. Distance runners have a tendency to train on much steeper grades than those that have ever been researched before, so it is important to understand how the GRF changes in these conditions. The absolute GRFs observed in the present study were much greater than those seen in previous studies; however, this was expected because of the steeper grades that were used. Prior studies also have either included a population of just males, or of males and females combined, while the present study analyzed the biomechanics of just females.

In the horizontal direction, it was expected that the braking forces would increase significantly, while the propulsive forces would decrease. Prior research examining the effect of downhill running on the braking force has been inconsistent. Gottschall and Kram (2005) and Chang and Kram (1999) found decreased propulsive forces and increased braking forces in downhill running compared to level running in their studies. In contrast, Telhan et al. (2010), and

Yokozawa et al. (2005) did not find differences in either of the horizontal forces. In the present study, the propulsive forces were significantly smaller as the grade declined, however, there were no changes in the braking force. The decrease in the propulsive force when going downhill may be due to the increased gravitational force. There is less of a need for a high propulsive force because the increased height increases the amount of potential energy that can then be transferred to kinetic energy. An increased braking force was expected, however, the decrease in the propulsive force may have eliminated the need for changes in the braking force.

In addition to increased vertical GRFs, there were also significantly greater loading rates seen in the present study. Both the average and the peak loading rates increased in a dose-response fashion as the grade became steeper. The average loading rate increased from 2.5 BW per second at level to almost 9 BW per second at -20%. This is an incredibly high rate of force development that the body must dissipate. This may be attributed to peak GRF shifting from the active peak to the impact peak when running on the steeper declines. These findings are similar to what has been seen in previous studies (Gimenez et al., 2014; Gottschall & Kram, 2005). However, no previous studies have established that there is a dose-response relationship between the grade and the loading rate. Previous research has indicated that the higher the loading rate is, the greater the risk for stress fractures becomes (Bennell et al., 2004; Burr et al., 1996). Therefore, the current study provides a strong indication that downhill running may significantly increase the risk of bone deformation.

The tibia and fibula are among the top locations that commonly sustain stress fractures, and track and field and distance runners are some of the most likely athletes to experience stress fractures (Brukner et al., 1996). Therefore, it is important to be aware of the potential risks of sustaining this kind of injury, especially during downhill running.

Joint Kinetics

Power absorption and production at the lower extremity joints.

Power absorption at each of the lower extremity joints increased significantly in the present study, while the power production at the knee and ankle significantly decreased. There is conflicting research on the power absorbed by the ankle joint. Buczek and Cavanagh (1990) found an increase in the power absorbed at the ankle, while Yokozawa et al. (2005) and Telhan et al. (2010) found no differences in the ankle power absorbed by the ankle. The present study showed a significant increase in ankle power absorption in each of the downhill grades compared to level, and the amount of power absorbed increased by almost half between level and -20%. This result is likely because the impact forces are transferred from the foot up through the body, which means the ankle is required to dissipate much larger forces in comparison to the knee and the hip (Zhang et al., 2000). In addition, the impact peak became much greater in the steeper grades, so there was a greater amount of force that needed to be absorbed.

The knee power absorption nearly doubled between the level condition and the -20% condition. The increased knee power absorption is consistent with previous research (Buczek & Cavanagh, 1990), however, the present study also established that there is a dose-response relationship between the decline and the amount of power absorption at the knee joint. This is thought to be the reason for the significant soreness felt by runners when training downhill (Telhan et al., 2010; Yokozawa et al., 2005). A significant increase in power absorbed by the hip in the steepest condition was also observed in the present study, although to a lesser extent than at the knee and ankle. This may be due to the ankle and knee dissipating a large portion of the forces before they reach the hip joint. Among the three joints, the knee absorbed considerably more power at each of the grades, followed by the ankle and then the hip. Additionally, the

difference in power absorption between the joints became increasingly greater as the decline became greater. This may be an indication that there is a greater risk for injury at the knee when running downhill compared to the ankle and hip.

The decrease in power production at the ankle and knee joints is likely related to the decrease in overall power propulsion needed in the horizontal direction. As mentioned before, this could be due to the steeper grades having greater potential energy, due to the runners landing from a greater height, that is then converted into kinetic energy. This means there is not as much power production needed by the lower extremity joints. Among the three joints, the ankle produced the most power at each of the grades, including level.

Joint moments in the sagittal plane.

The sagittal plane joint moments observed partially supported our hypothesis. There was an increase in the peak plantar flexor moment in the -5% and -10% conditions compared to level, however, at -15% and -20% the peak plantar flexor moment was smaller than in the level condition, which was not expected. However, the decreased plantar flexor moment does correlate to the decrease seen in the medial gastrocnemius muscle activity. Previous research has shown a decreased plantar flexor moment when running downhill (Yokozawa et al., 2005), which is similar to our results, or no differences at all in the ankle joint moments (Telhan et al., 2010). There were no significant differences in the dorsiflexion moment between the downhill conditions and the level condition. This is contrary to what was expected, however, it does correlate to the lack of changes seen in the tibialis anterior muscle activity.

The knee extension moment during the level condition was greater than in the -5% condition, but there were no other significant differences between the downhill grades when compared to the level condition. There also were no differences seen in the knee flexion

moment. This is similar to what has been found in the past (Buczek & Cavanagh, 1990), however, it was still unexpected because the muscle soreness that is often reported by runners is generally in the quadriceps muscles, or the knee extensors. Although previous studies had reported no changes in the knee extension moment (Buczek & Cavanagh, 1990; Telhan et al., 2010), we expected to see an increase because the grades that the present study used were so much greater. The lack of differences seen in the knee joint moments do, however, correlate to the absence of changes seen in the biceps femoris and vastus medialis muscle activity.

In contrast to other studies, the peak hip extension moment in the present study was significantly greater in each of the downhill grades compared to the level condition. Yokozawa et al. (2005) found a decrease in the peak hip extension moment, and Telhan et al. (2010) found no change at all. The increase in the hip extension moment along with the decreases in the plantar flexion moment indicates that the hip muscles may be contributing more work when running downhill at steeper grades.

Joint moments in the frontal plane.

There were no significant differences in the peak ankle eversion moment, but there was a significant increase in the peak ankle inversion moment in each of the downhill grades compared to level. It is interesting that the inversion moment increased, because the peak ankle eversion angle decreased in the downhill conditions. Individuals with ITBFS have been found to have a significantly greater rear foot inversion moment compared to healthy runners (Ferber et al., 2010), which is an indication that this variable may contribute to an increased risk for sustaining ITBFS when running downhill.

The knee adduction moment increased significantly in each of the downhill grades compared to level, and there was also a dose-response relationship. Although there were no

significant differences in the frontal plane angles at the knee, the increased adduction moment indicates increased lateral stress placed on the knee. The increased stress on the lateral aspect of the knee may lead to lateral knee pain over time, which is a common complaint with ITBFS.

There was an increased adduction moment at the hip in the -15% and -20% grades, which correlates to the increased adduction moment seen at the knee. The increased hip adduction moment may lead to overuse of the hip adductor muscles. This in turn may cause fatigue and muscle soreness in the hip adductor muscles. However, the increased hip adduction moment could also result in the hip adductor muscles becoming stronger, which would be a benefit of downhill training.

Joint Kinematics

Joint motion in the sagittal plane.

The kinematics utilized in running have the ability to affect the magnitude of the impact shock felt. Previous research has shown that downhill running results in a greater angle at the shank, greater knee extension, and less hip flexion at initial contact (Paradisis & Cooke, 2001). The results of the present study are in line with these findings. Our results showed decreased knee and hip flexion at initial contact during downhill running, which may lead to greater impact forces. There was also a greater peak knee flexion angle seen. The results seen at the knee are identical to what was found by Buczek and Cavanagh (1990) and Mizrahi, Verbitsky, and Isakov (2000). However, both of these previous studies used only male runners, and used much smaller declines, but the matching results show that downhill running will cause the knee kinematics to change. The decreased knee flexion at initial contact is likely because the runners have to extend further in order to reach for the lower ground (Yokozawa et al., 2005), and the greater ROM at the knee has been attributed to needing to cushion the landing more (Buczek & Cavanagh, 1990).

Additionally, the present study also established a dose response relationship between the downhill grade and the resulting ROM at each of the lower extremity joints. As the downhill grade became steeper, the ROM at both the ankle and the hip decreased, and the ROM at the knee increased. Additionally, there was a greater amount of trunk extension when running downhill, which is similar to previous research (Gimenez et al., 2014). This means that runners are leaning backward more when running downhill, which may also increase the magnitude of the GRFs.

The decreased ROM seen at the ankle and hip in this study indicates stiffer joints, which may contribute to the increased GRFs and the increased loading rates observed in this study. The decreased ROM at the ankle might also be linked to the decreased power production seen at the ankle, the decreased plantarflexion moment, and the decreased muscle activity seen in the medial gastrocnemius muscle. The increased ROM at the knee seen when running downhill could also contribute to ITBFS and PFPS in distance runners. In runners with ITBFS, the posterior edge of the IT band impinges against the lateral femoral epicondyle shortly after initial contact. The impingement generally occurs around 20 to 30 degrees of knee flexion and repetitive irritation can lead to increased inflammation (Orchard et al., 1996). Runners have reported that running downhill increases irritation of the condition (Fredericson & Wolf, 2005), which may be because downhill running involves landing in a more extended position and moving through a greater ROM in the stance phase (Hamill et al., 2008), which was observed in the present study. Anterior knee pain, as occurs with PFPS, is also heightened by increased flexion and extension at the knee (Rolf, 1995), so downhill running may also increase the risk of PFPS in runners.

Joint motion in the frontal and transverse planes.

There were decreased ankle eversion angles seen in the downhill grades. Ankle eversion is one of the three combination movements involved in pronation. Pronation allows for forces to be attenuated over a longer period of time, and thus serves as a protective mechanism during running to absorb the high GRFs (Hreljac & Ferber, 2006; Subotnick, 1985). As the grade became steeper, the eversion angle decreased, and the ankle became stiffer. This is an indication that the forces were not being attenuated as well, and may increase the risk for injuries such as stress fractures.

The IT band originates on the iliac crest and terminates on the fibular head, passing over the lateral femoral condyle. Previous research has shown that females who develop ITBFS, or who have a history of ITBFS, exhibit greater femoral external rotation when running (Noehren, Davis, & Hamill, 2007). Results of the present study showed increased hip external rotation in the steeper declines, and the peak internal rotation also decreased, so the hip was in a greater amount of external rotation throughout stance. Greater hip external rotation, without a change in knee external rotation means the runners were landing more on the lateral side of the foot, which correlates with the decreased eversion observed. Greater external rotation at the hip will place increased stress on the distal end of the IT band, which increases the risk for ITBFS. No previous research has reported frontal or transverse plane motion when running downhill, therefore, the results from this study may serve as a baseline for future research in this area.

Spatiotemporal Parameters

There is an inverse relationship between the stride rate and step length, horizontal distance between the COM and the heel at initial contact, and the braking force (Heiderscheit et al., 2011). To maintain the same speed, if the stride rate increases, the step length will decrease,

and the heel is placed closer to the COM, which then reduces the braking impulse. This also results in less power being absorbed at the knee and hip.

It was hypothesized that as the downhill grade became steeper that the stride rate would decrease and the step length would increase, which would result in a greater braking force being observed, as well as increased knee and hip power absorption. The present study showed no changes in either the stride rate or the step length. This could help to explain why there were no changes in the braking force observed in this study. Previous research showed conflicting results regarding these spatiotemporal parameters. One study found a decreased stride rate (Gimenez et al., 2014), while others found an increased stride length during downhill running (Paradis & Cooke, 2001), and still others found no differences at all in the stride rate or the step length (Chang & Kram, 1999; Yokozawa et al., 2005). Although the grade was expected to cause changes in the stride parameters, the participants were all trained distance runners and maintained very consistent stride patterns across all of the conditions. Since the speed remained the same, there was no need for the runners to alter their stride.

Electromyography

Previous research indicated that the most active muscles during downhill running are the gluteal muscles, the quadriceps, and the calves, which are the extensor muscles of the lower extremity (Buczek & Cavanagh, 1990). The decreased hip and knee flexion seen in downhill running would be expected to result in a greater amount of muscle activation in the extensor muscles. Downhill running has also been found to be connected to the greatest onset of DOMS compared to level and uphill running, likely due to an increase in eccentric muscle activity that is required when running downhill (Brown et al., 1999; Cai et al., 2010).

It was hypothesized that there would be changes in the level of muscle activation when running downhill, however, it was unpredictable as to what those changes would be because of conflicting results in past research. However, the only differences observed occurred in the medial gastrocnemius muscle, and the observed muscle activity decreased. This was a surprising finding, however, it is consistent with the other findings from this study. The decreased ROM at the ankle and the decreased plantarflexion moment, both may be related to the decrease in gastrocnemius activity that was seen. Additionally, the more rigid ankle joint could be linked to the decrease in power production at the ankle, which would also explain the decrease in muscle activity seen in the gastrocnemius muscle. Furthermore, there were no differences seen in the in hip adduction when running downhill, which explains why there were no changes in the muscle activity seen in the gluteus medius muscle, as this muscle acts as a hip stabilizer. Although there were no changes in the muscle activity between the different conditions, the vastus medialis muscle was the most active compared to the other muscles analyzed, which is consistent with past research (Buczek & Cavanagh, 1990). While more differences were expected, the findings are consistent with other results from this study.

Conclusions

The primary goal of this study was to quantify the differences in joint mechanics, muscle activity, and spatiotemporal patterns among different grades, and to identify if a dose-response relationship existed, as well as to determine if the risk for lower extremity injury increases when running downhill. Previous research on downhill running has been limited, and there are many conflicting results in the studies that have been done. The present study is, to the best of our knowledge, the first study to examine downhill running on grades similar to those that would be encountered on long trail runs, and to analyze the biomechanics of just female runners. It is also

believed to be the first study that has examined whether a dose-response relationship exists between the grade and each variable studied.

The most significant findings of this study were the greater vertical GRFs, increased loading rates, increased power absorption at the lower extremity joints, and changes in the ROM at the lower extremity joints seen with downhill running. Increased GRFs and loading rates were expected, however, the extent to which the loading rates increased was much greater than anticipated. The differences in the GRFs and the loading rates when running downhill compared to running on a level surface are strong indicators for potential stress fractures, as well as for PFPS. The increased ROM at the knee and the decreased knee flexion at initial contact are risk factors for knee pain such as in PFPS and ITBFS. Additionally, the decreased knee and hip flexion and trunk extension at initial contact may contribute to the increased GRF and loading rate in the steeper grades. Overall, there appears to be a greater risk for sustaining an injury when running downhill. However, this was not a prospective study so it cannot be stated that downhill running causes injuries. Additionally, hill running, up and down, is a key part of distance training (Paradisis et al., 2009). Therefore, hill running should not be cut out of the program entirely, but rather knowledge of the effect that downhill running has on the biomechanics, and the resulting risk for injury, should be kept in mind so that training programs may be modified as necessary in order to avoid overuse injury.

Future Recommendations

The present study only used one running speed in order to isolate grade as the independent variable. Future research should take into account how different running speeds may affect the biomechanics of downhill running. Additionally, differences in preferred foot strike pattern might also affect the biomechanics of downhill running.

Chapter 7

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Appendix A

Assess your health status by marking all *true* statements

History

You have had:

- ☐ a heart attack
- ☐ heart surgery
- ☐ cardiac catheterization
- ☐ coronary angioplasty (PTCA)
- ☐ pacemaker/implantable cardiac defibrillator/rhythm disturbance
- ☐ heart valve disease
- ☐ heart failure
- ☐ heart transplantation
- ☐ congenital heart disease

If you marked any of these statements in this section, consult your physician or other appropriate health care provider before engaging in exercise. You may need to use a facility with a medically qualified staff.

Symptoms

- ☐ You experience chest discomfort with exertion
- ☐ You experience unreasonable breathlessness
- ☐ You experience dizziness, fainting, or blackouts
- ☐ You take heart medications.

Other health issues

- ☐ You have diabetes
- ☐ You have asthma or other lung disease
- ☐ You have burning or cramping sensation in your lower legs when walking short distances
- ☐ You have musculoskeletal problems that limit your physical activity
- ☐ You have concerns about the safety of exercise
- ☐ You take prescription medications
- ☐ You are pregnant

Cardiovascular risk factors

- ☐ You are a man older than 45 years
- ☐ You are a woman older than 55 years, have had a hysterectomy, or are postmenopausal
- ☐ You smoke, or quit smoking within the previous 6 months
- ☐ Your blood pressure is >140/90 mm Hg
- ☐ You do not know your blood pressure
- ☐ You take blood pressure medication
- ☐ Your blood cholesterol level is >200 mg/dL
- ☐ You do not know your cholesterol level
- ☐ You have a close blood relative who had a heart attack or heart surgery before age 55 (father or brother) or age 65 (mother or sister)
- ☐ You are physically inactive (i.e., you get <30 minutes of physical activity on at least 3 days per week)
- ☐ You are >20 pounds overweight

If you marked two or more of the statements in this section you should consult your physician or other appropriate health care provider before engaging in exercise. You might benefit from using a facility with a professionally qualified exercise staff^a to guide your exercise program.

- ☐ None of the above

You should be able to exercise safely without consulting your physician or other appropriate health care provider in a self-guided program or almost any facility that meets your exercise program needs.

^aProfessionally qualified exercise staff refers to appropriately trained individuals who possess academic training, practical and clinical knowledge, skills, and abilities commensurate with the credentials defined in Appendix D.

FIGURE 2.2. AHA/ACSM Health/Fitness Facility Preparticipation Screening Questionnaire (Modified from American College of Sports Medicine Position Stand and American Heart Association. Recommendations for cardiovascular screening, staffing, and emergency policies at health/fitness facilities. *Med Sci Sports Exerc.* 1998;30(6):1009–18.)

Health/Running Information

Ball State University – Biomechanics Laboratory

Health/Running Information

Subject ID _____

Gender: Male ____ Female ____

Age: ____ Height: _____ Weight: _____

Whom to contact in a case of emergency _____ Ph# _____

Name of your physician _____ Ph# _____

1. Have you ever been diagnosed as having any of the following conditions?

Yes _____ No _____ If yes, please put approximate year of onset in space provided.

Heart attack _____

Angina (chest pain) _____

Respiratory disease _____

Neuropathies _____

Other neurological conditions _____

Osteoporosis _____

Rheumatoid arthritis _____

Other arthritic conditions _____

Other movement disorders _____

2. Have you ever been diagnosed as having any of the following conditions?

Yes _____ No _____ If yes, please describe what kind.

Joint replacement _____

Uncorrected visual problems _____

Other health problem? _____

3. Do you currently suffer any of the following symptoms in your legs or feet? Please check the space of all that apply.

Numbness _____ Tingling _____ Arthritis _____ Swelling _____

4. List all medications that you currently take (including ‘over-the-counter’ medications)

Type of Medication

For what condition

_____	_____
_____	_____
_____	_____
_____	_____

5. Have you required emergency medical care or hospitalization in the last three years?

Yes _____ **No** _____ **If yes, please list when this occurred and briefly explain why.**

6. Have you ever had any condition or suffered any injury that has affected you balance or ability to walk without assistance?

Yes _____ **No** _____ **If yes, please list when this occurred and briefly explain condition or injury.** _____

7. How would you describe your health?

Excellent _____ Very good _____ Good _____ Fair _____ Poor _____

Running History

1. How many days per week do you exercise?

One ____ Two ____ Three ____ Four ____ Five ____ Six ____ Seven ____

2. How many days per week do you run?

One ____ Two ____ Three ____ Four ____ Five ____ Six ____ Seven ____

3. How many miles do you run each day? _____

4. How many miles do you run per/ week? _____

5. What average pace do you run these miles? _____

6. How long have you been running at this volume? _____

7. How many years have you been running? _____

8. Do you have any experience running hills? Yes _____ No _____

If yes, explain _____

9. Do you have any experience running on trails? Yes _____ No _____

If yes, explain _____

10. Have you had a running related injury in the past 2 years? Yes _____ or No _____

If yes, please list when this occurred and briefly explain condition or injury _____
